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15 January 1953

From: Commander, U. S. Naval Air Development Center, Johnsville, Pa.
To: Chief, Bureau of Medicine and Surgery

Subj: Report No. NM 001 060.07.01 TED NADC AE-1401.06, and Office of Naval Research, Contract N6-ONR-249, Task Order 4, Phase I, titled "The Evaluation of Pressure Transducer Systems," of BuMed Project No. NM 001 060; forwarding of

Ref: (a) BuMed Form 98 approved 16 January 1951
(b) BuMed ltr of 17 January 1951

Encl: (1) Report No. NM 001 060.07.01 TED NADC AE-1401.06 "The Evaluation of Pressure Transducer Systems"

1. This study report is prepared in compliance with references (a) and (b) and is herewith forwarded as a first report of the subject under the BuMed Project No. 001 060 titled "Acceleration and Deceleration Studies With the Human Centrifuge and Research Aircraft."

Robert S. Hatcher

Robert S. Hatcher

Copy to:
Chief, Bureau of Aeronautics

AVIATION MEDICAL ACCELERATION LABORATORY

NAVAL AIR DEVELOPMENT CENTER

JOHNSVILLE, PENNSYLVANIA

REPORT NO. NADC-MA-5206

U. S. NAVAL AIR DEVELOPMENT CENTER

AVIATION MEDICAL ACCELERATION LABORATORY

JOHNSVILLE, PENNSYLVANIA

DEVELOPMENT OF BIOLOGICAL RESEARCH APPARATUS FOR USE IN ACCELERATION AND DECELERATION STUDIES

Phase I

The Evaluation of Pressure Transducer Systems

Bureau of Medicine and Surgery
Report No. NM 001 060.07.01
Study No. NM 001 060.07
of
Project No. NM 001 060
and
Office of Naval Research Contract N6-ONR-249, Task Order 4

Reported by:

John Parnell
John Parnell, M.S. in E. E.
Moore School of Electrical Engineering
University of Pennsylvania

Edw. L. Beckman
LCDR Edw. L. Beckman MC USN
Aviation Medical Acceleration Laboratory, and
Associate in Physiology, School of Medicine
University of Pennsylvania

Lysle H. Peterson, M. D.
Assistant Professor of Physiology, School of Medicine
University of Pennsylvania
Lysle H. Peterson

Approved and Released by:

L. D. Carson
CAPT L. D. Carson MC USN
Director
Aviation Medical Acceleration Laboratory

Aviation Medical Acceleration Laboratory

REPORT NO. NADC-MA-5206

NOTE

This investigation was carried out jointly by the personnel of the Electromedical Research Laboratory, Moore School of Electrical Engineering, University of Pennsylvania; Department of Physiology, School of Medicine, University of Pennsylvania; and the Aviation Medical Acceleration Laboratory, Naval Air Development Center. The portion of the work carried out by personnel of the University of Pennsylvania was supported by research grant from the Office of Naval Research Contract N6-ONR-249, Task Order 4. Absolute calibration of the experimental apparatus was made by the personnel of the Underwater Sound Reference Laboratory, Office of Naval Research, Orlando, Florida.

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Aviation Medical Acceleration Laboratory

REPORT NO. NADC-MA-5206

SUMMARY

In Phase I of this investigation, comparison measurements were made of the electrical output of various commercially available pressure transducers. These transducer systems were tested utilizing various coupling systems (catheters and needle combinations) which are currently being used for physiological measurement. The electrical output of these systems was measured in two ways: (a) when the input pressure was applied by a variable frequency sine wave pressure pump (pistonphone), and (b) when the amplitude versus frequency response was calculated from the measured response of the system to a pressure step function. Measurements obtained in this manner were then compared with the amplitude versus frequency response of these same systems as measured by absolute calibration methods. The facilities of the Navy Underwater Sound Reference Laboratory, Orlando, Florida, were used to obtain absolute response measurements. Graphs showing the amplitude versus frequency response of the various systems measured as well as the comparison curves to demonstrate the reliability of the conventional methods of response measurement are included as Appendix B.

In Phase II of this investigation, the theory of transducer response for lumped systems (considered in Phase I) was extended to a consideration of a transducer system having more than one degree of freedom (distributed system). For this purpose, the capacitance type transducer with a very long polyvinyl catheter was used as the test system. Measurement of physical constants of this system were carried out and the electrical transmission line theory was applied to the catheter. This theory was then tested for closeness of fit by comparing the measured response of the system to the response of the system calculated in accordance with the developed theory. Also, theoretical consideration was given to the effect of varying physical constants upon the response of the system. These effects are shown in graphs included in the text of this report. A damping segment was designed based upon the theoretical concepts. This unit was found satisfactory for producing variable damping in a transducer system, and, thereby, improving the amplitude versus frequency response of the system and reducing acceleration artifacts.

INTRODUCTION

A. PURPOSE

In this study of the circulatory system, it is desirable to be able to measure relatively small pressures which are instantaneously varying with time. Such measurements are particularly significant in physiological investigations on the human centrifuge where very rapid changes in pressure are observed. It is further desirable to make these measurements with minimal alteration of the flow characteristics and of the function of the circulatory system. Devices for converting a rapidly varying pressure phenomenon into an electrical replica have been termed transducers. In order to obtain pressure measurements with minimal alteration of the circulatory system, investigators have found that either the transducer must be of relatively small diameter with respect to the blood vessel so that it may be introduced directly into the blood vessel in which the pressure is to be measured, or the transducer must be connected to the point of measurement through a hydraulic transmission line, or catheter, which may be inserted into the vessel to the point of measurement (1,2,3, and 4). Technical difficulties have limited the development of a

generally satisfactory system of the first type. However, systems of the second type, utilizing a small relatively stiff catheter as a hydraulic transmission line, have been developed which are quite successful.

B. IMPORTANCE OF PROBLEM

Although technical developments in the past few years have made available several types of pressure transducer systems which have been utilized in cardiovascular research, research and development into methods of testing these devices have been relatively limited. The acceptability of pressure transducer systems for physiological use has primarily been based upon the measurement of the frequency response of the system to a sine wave function of pressure or to a square wave function of pressure. Evaluation of transducer systems by their response to a variable frequency pressure sine wave is generally accomplished by the measurement of the system output when the input pressure is supplied by a variable frequency sine wave pressure pump, frequently called a pistonphone. In order to utilize a pressure pump as a measuring system, it is apparent that some method of calibration of the pressure pump is necessary in order to

adequately evaluate the system being tested. When a transducer system is evaluated on the basis of its response to a square wave pressure function, the output amplitude versus frequency response is determined by measurement of the oscillations of the system when excited by a pressure step function and the application of a series of mathematical calculations presented by Frank (5) to these measurements. However, these calculations assume that the transducer system may be treated as a system with one degree of freedom consisting of a concentrated mass, a spring, and a constant friction force. If visco-elastic catheters are used as fluid transmission lines, (an application for physiologists), this assumption of a simple mechanical system may not be valid. An evaluation of the validity of these assumptions when applied to transducer systems being used in physiological research is, therefore, indicated.

C. APPROACH TO PROBLEM

Before discussion of this problem, it is desirable to define the following terminology:

The system refers to the complete hydraulic path from the site of pressure measurement to the point of conversion

of mechanical to electrical energy.

A simple system (or system of one degree of freedom) is one which has only one appreciable mass, spring, and friction loss such as

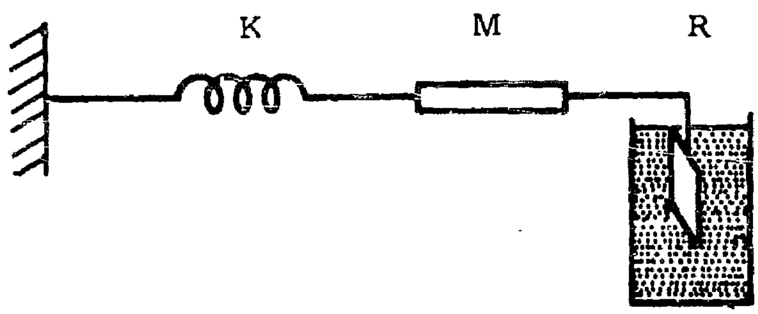


Figure 1 - Diagram of Simple Mechanical System.

A distributed system is one which has a large (or infinite) number of masses, springs, and friction losses as

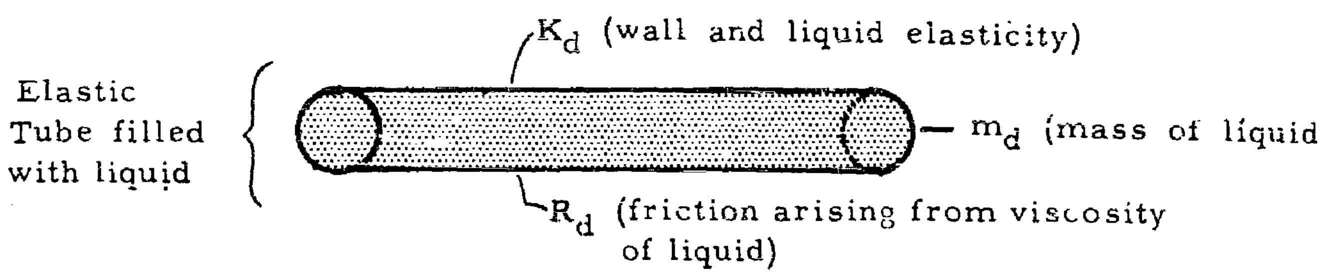


Figure 2 - Diagram of Distributed Mechanical System.

The lumped system is often used interchangeable with a simple system.

Response is used to denote the electrical output of a

transducer corresponding to a defined pressure input function.

Step function response means the shape of the output voltage-time curve as a sudden change from one pressure to another is impressed on the input.

Amplitude-frequency response is the shape of the envelope of the output voltage-time curve when a pressure sine wave is impressed on the input, and the frequency of the wave is varied uniformly with time from a low to a high value.

Resonance is the term applied to any point on the amplitude-frequency response where the curve goes through a maximum value. In a distributed system, many resonances occur corresponding to the frequencies where the system is a multiple of a half wave length.

The first resonance mode is the frequency at which the system is a half wave length; the

second resonance mode is the frequency at which the system is a full wave length, and so on.

Most of the mathematics used in this paper are taken from

electric circuit theory; i.e., certain mechanical assemblies are recognized as being governed by the same differential equations as specific electrical configurations; the only differences are in the meaning of the constants and quantities differentiated. This similarity of equations leads to the concept of analogs, or the relation between an electrical quantity and its corresponding mechanical quantity in their corresponding mathematics. For example, the simple system of Figure 1 is the analog of the series resonant circuit.

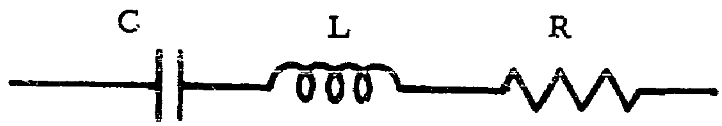


Figure 3 - Diagram of Electrical Analog of Simple Mechanical System.

The mass m is the analog of the electrical inductance L; the spring constant K is the analog of the reciprocal of the capacitance $^1/C$; and the friction constant R is the analog of the resistance R_e. Similarly, the distributed system of Figure 2 is assumed to be the analog of an

electrical transmission line.

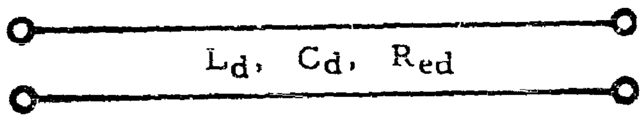


Figure 4 - Diagram of Electrical Analog of Distributed Mechanical System.

Justification of this assumption is dependent upon the agreement of the responses of the analogous systems, and, therefore, is considered later in the text of this report. However, the mass, spring constant, and friction constant, all per unit length, (m_d, K_d, and R_d) are the analogs of the inductance, reciprocal of capacitance, and resistance, all per unit length, (L_d, $^1/C_d$, and R_{ed}). It is to be noted that the leakage conductance, per unit length C_d has been omitted, but will be considered later. Other analogs necessary to complete the relation between the distributed electrical and mechanical systems are for the current, the voltage, and the charge (i, v, and q). The analogs of these quantities are velocity, force, and position, respectively (V, F, and x). Definitions for all symbols used herein are included in Appendix A.

The purpose of these investigations is to compare the response characteristics of several physiological pressure transducer systems as determined by conventional methods (response to a pressure step function and the amplitude-frequency response to a pressure sine wave) with the response characteristics as obtained by using absolute calibration methods.* Based upon these absolute calibrations, the approximate theory for distributed systems as postulated by Hansen (6) has been examined on the basis of electrical transmission line theory. The correlation of the fit between the electrical analog and the distributed mechanical system has been tested by comparing the theoretical with the absolute response characteristics. It is felt that a better understanding of the theory of transducer systems of both the simple and distributed types would permit a more effective utilization of this type of apparatus by physiologists engaged in acceleration research at the Aviation Medical Acceleration Laboratory. Therefore, in Phase II of this report, the theory of a distributed system is considered and a formulation of a mathematical expression for the response

* Absolute calibration of the experimental apparatus was made by the personnel of the Office of Naval Research, Underwater Sound Reference Laboratory, Orlando, Florida.

of such a system is derived and tested against experimental data. Various physical factors of the system are varied and the theoretical effect of such variations calculated. Based upon the implications of these derivations, a damping device has been designed which would be effective in decreasing the resonance frequency overshoot, thus making the transducer system a more linear measuring instrument.

REVIEW OF THE LITERATURE

The study of transducers was first undertaken by physicists working in the fields of physics and engineering of acoustics and hydroacoustics. Considerable progress has been made with the theory as well as the use of transducers.

The application of acoustical knowledge to the specific problems of measuring physiological pressures was first reported by Frank (5) in which he presented a theoretical and experimental analysis of transducer systems in which component sections responded to pressure phenomena, simultaneously. Frank used a rapid change in static pressure applied to the transducer input as a test signal. The response or output of the transducer was used as a basis upon which to analyze the performance of the system. He recognized this response as the solution to a particular differential equation and applied numerous mathematical techniques to this equation, so that the manner in which the output amplitude varied with frequency could be calculated when a constant amplitude-variable frequency sinusoidal pressure was applied to the transducer output. Based upon the principles of manometry elaborated by Frank (5), numerous investigators undertook the design

of physiologically useful pressure transducers utilizing various different physical principles. Wiggers (7), Broemser (8), and Hamilton (9), developed mechanico-optical transducer systems in which the pressure variation was used to cause variable flexion of a glass or metal diaphragm, in which the amount of flexion of the diaphragm was then optically amplified, enabling the signal to be calibrated against known pressures. Many mechanico-electrical transducers have been developed utilizing various electrical variables as indicating systems. Variable resistance systems have been developed by Schütz (10) and by Wagner (11). Lambert and Wood (12) have successfully modified a commercially available variable resistance transducer for physiological use. The use of the piezo-electric characteristics of certain crystals (quartz, and barium titanate) has been investigated as the basis for a physiological pressure transducer without remarkable success (MacLeod and Cohn (13) and Lowry (14)). However, investigations into the use of variable inductance transducer systems have been more successful. This system, devised by Wetterer (15) and later modified by Gauer (1) should prove a very useful physiological tool.

The use of a variable capacitance circuit in which the distance between condenser plates is varied by variations in pressure has been investigated by several experimenters, including Schütz (16); a group of Danish workers —Bucthal and Warburg (17), Skouby (18), and Hansen (6); and by an American group Lilly (19), Peterson, Dripps, and Risman (3). Perhaps the most significant contribution was made by the Danish group. They improved upon the equipment of Frank and extended the theory to include systems which did not include oscillations when excited; i.e., critically damped and overdamped systems. These investigators employed a variation of the pistonphone method of response measurement and compared the pistonphone results with those which were calculated from a rapid change in static pressure. They were also the first experimenters to propose an approximate theory for systems in which the effects of a varying pressure phenomena did not occur simultaneously (distributed systems). This condition exists when plastic catheters are used. A presentation of the results of their findings together with the techniques of use are presented in an English language monograph (Hansen (6)), along with a very complete review of the development of the theory and techniques of manometry.

Concomitant with the development of physiologically useful pressure transducers by physiologists and biophysicists, investigators interested in hydroacoustics were responsible for developments which were valuable in the field of physiological manometry—the extension of the theorem of reciprocity to the calibration of acoustic transducers (20), and the development of absolute low frequency calibration system for hydrophones (21). These two contributions in hydroacoustics were significant because they made available a system of measurement which did not require the use of a standard of assumed calibration.

EXPERIMENTAL PROGRAM

This paper has been divided into two phases. In Phase I, the theory of transducer system response for simple systems will be examined.

Response measurements were made of various pressure transducers and transducer fittings and coupling systems using pressure step function, sine-wave pressure, and absolute calibration techniques. The results of these tests will be compared to determine the adequacy of conventional testing procedures and thereby to evaluate the assumption that these systems may be treated as simple or lumped systems.

In Phase II, the theory for transducer systems which was tested in Phase I will be expanded to consider systems containing a relatively elastic catheter and one which is relatively long in relation to the half wave length of the frequency to be studied. This theory will be tested experimentally and will be examined by variation of some of the equation coefficients.

PHASE I

LUMPED SYSTEMS

A. DEVELOPMENT OF EVALUATION METHODS; THEORY OF LUMPED SYSTEMS

When this work was first undertaken, some time was spent seeking a method of determining the ability of a transducer system that would reproduce a rapid pressure variation. There were two possibilities open: (a) the measurement of the response of a system to a pressure step function and subsequent calculation of the amplitude versus frequency response, and (b) the measurement of the amplitude of the output signal when a variable frequency pressure sine wave of constant amplitude was impressed upon the system. Because the step function could be produced more readily, and because it represented the phenomena being measured more accurately than a sine wave, it was chosen for the initial tests. Earlier workers produced pressure step functions in the following ways: (a) hydraulic pressure was suddenly applied to the transducer by rapidly opening a stopcock to a connected pressure source, (b) air under pressure was applied to a chamber over the transducer and then suddenly released by rupturing

a paper or rubber diaphragm, (c) hydraulic pressure was rapidly decreased by pulling back on a fluid filled syringe coupled to the transducer, or (d) a hydraulic pressure on the transducer was decreased by sucking on the coupling needle or catheter, then the pressure orifice was sealed with the tongue or finger which was then rapidly removed from the orifice releasing the pressure. The latter system was most satisfactory from the practical point of view. Method (b) was tried and found to produce extraneous vibration in addition to the step function. Pressure transducers alone are usually sensitive to movement (acceleration) as well as pressure, and transducers with fluid filled catheters are always very sensitive to movement. These extraneous vibrations obscured the step function response in this method and make it unsuitable. A variation of this method (b), the breaking of a glass tube, was tried and discarded for the same reason. The syringe plunger and stopcock methods were not tried because such equipment introduced a significant volume which could resonate. (The early tests were made on the transducer alone, and on the transducer with a stopcock and needle--both of these combinations have relatively high natural frequencies, i.e., 1200 cps.) A method was finally devised whereby a pressure step function could be

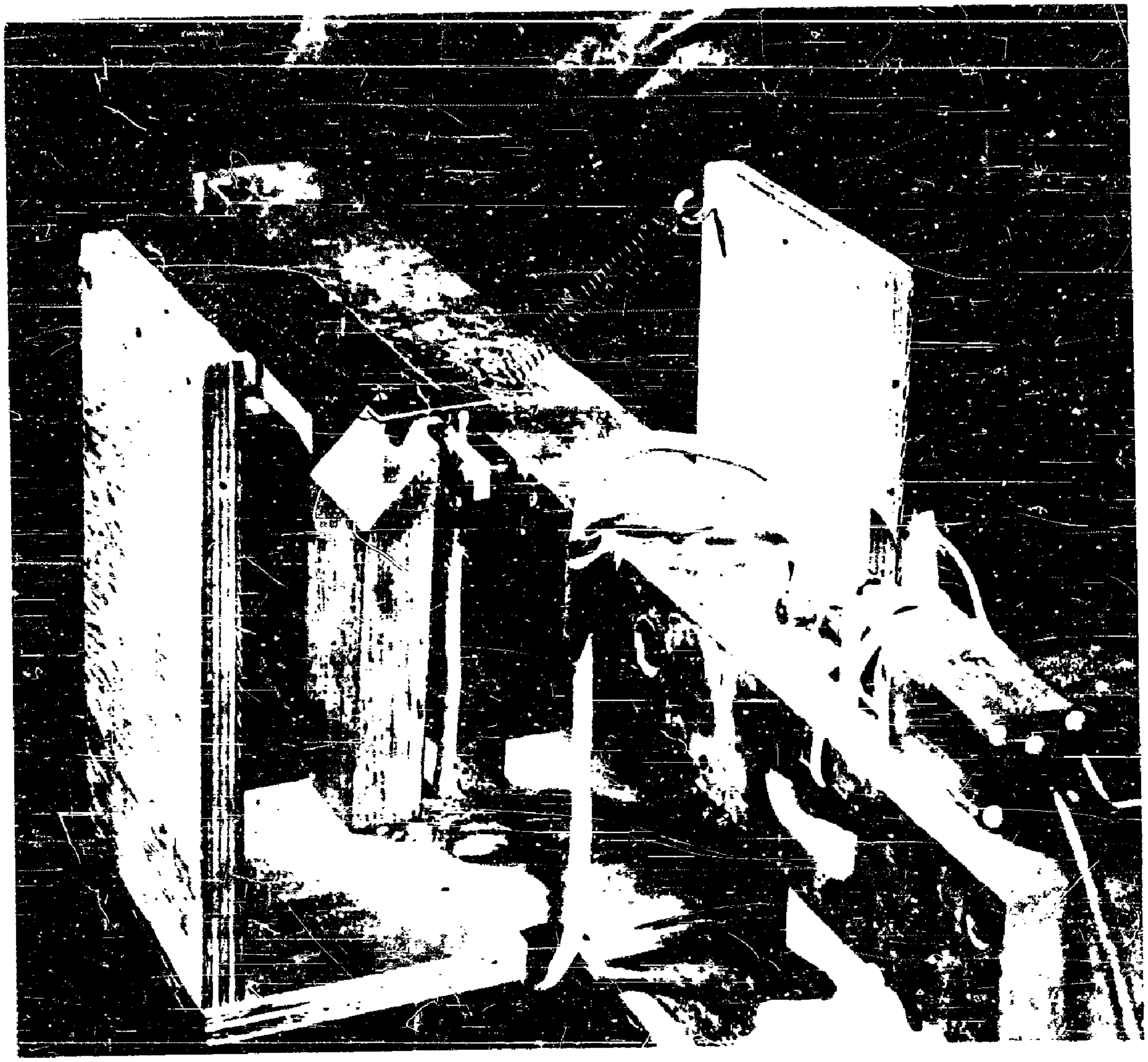


Figure 5 - Pressure Step Function Generator Showing Statham Transducer Being Tested.

produced with very little energy expenditure. It consisted of a 10 x 20 x 1 mm. plastic chip covered with rubber and suspended from two springs. This chip was held over the transducer opening by pressure from the experimenter's fingernails or by a thread from which was suspended a weight. After hydraulic pressure was applied from an external source, the chip was suddenly released by moving the fingers slightly or by cutting the supporting thread (Figure 5). The springs retracted the chip which opened the transducer to atmospheric pressure in a short, but measurable time interval. A plain plastic chip was also used to produce an impulse of negative pressure when a pulse with relatively short rise time or low amplitude was needed. (An impulse may be defined as a pulse of unspecified shape which, however, is unidirectional, and which has a duration much shorter than the duration of a desired response.)

A modification of system (d) was found useful for test purposes, such as the determination of whether or not a system was completely liquid filled. The plastic and rubber chip had one disadvantage in that the surface tension of the liquid caused a spurious pressure decrease when the chip was first released. Thus, the pressure function impressed

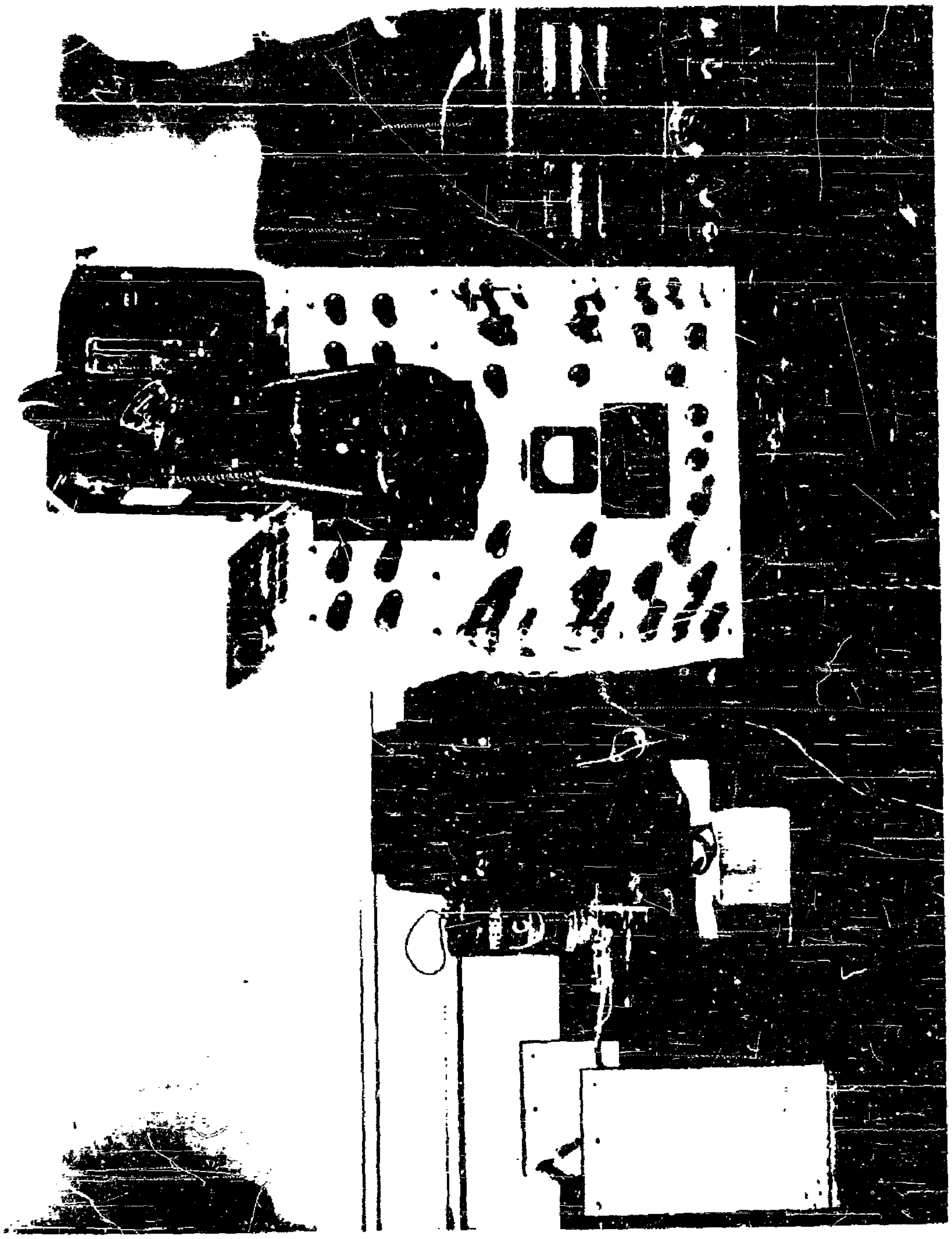


Figure 6 - Pressure Step Function Generator and Recording Equipment.

on the system under test was a combination of an impulse and a step function. Several transducers, many combinations of transducer and fittings, and combinations of transducer, fittings, and catheter were excited by the above described pressure functions, and records of their responses were made by the use of a dual beam cathode ray oscilloscope with d.c. amplification. The beam oscillations were recorded photographically by a camera with continuously moving 35 mm. film (Figures 6, 7, and 8).

B. RESULTS OF SYSTEMS MEASUREMENT

Although the transient response of a transducer system is a useful evaluation of its performance, it is also useful to consider the amplitude versus frequency response of the system. Engineers have used this response for communication system evaluation for many years and many pressure measuring systems used in physiology have been judged entirely from this response. In the case of the simple lumped system (system with one degree of freedom), it is a relatively simple matter to transform the step or impulse function response into an amplitude versus frequency response (6, 22, 23, and 24). The equation governing a simple, lumped, series resonant circuit driven by a sine

wave of constant amplitude and frequency is

$$m\frac{d^2x}{dt^2} + R\frac{dx}{dt} + KX = Y\sin \nu t \,. \qquad (1)$$

Where t = time

x = position of the mass m at the transducer opening,

R = mechanical resistance referred to the transducer opening,

K = spring constant (in Hook's Law) at the transducer opening,

Y = the amplitude of the driving force, and

ν = frequency of the driving force.

The steady state solution of this equation is

$$X(t)_{ss} = \frac{Y}{m\sqrt{\left(\frac{k}{m}-\nu^2\right)^2+\left(\frac{R}{m}\nu\right)^2}} \cos(\nu t+\phi)$$

where $\tan \phi = \frac{k}{m}-\nu^2 \Big/ \frac{R}{m}\nu$ (2)

The coefficient of the cosine indicates how the amplitude of the steady state driven system varies with the driving frequency ν for a fixed

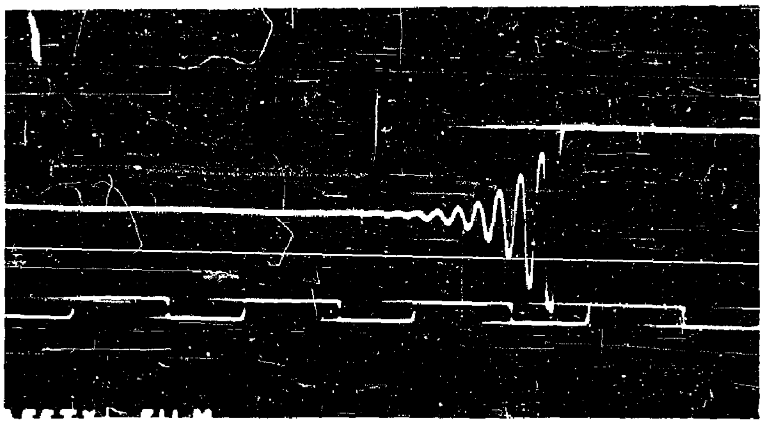


Figure 7 - Oscillogram of Step Function Response of Technitrol "Lilly" Transducer, Stopcock and Short No. 24 Needle.

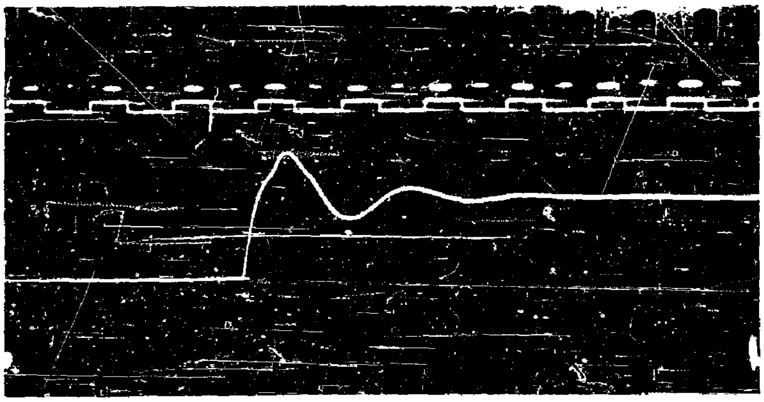


Figure 8 - Oscillogram of Step Function Response of Technitrol "Lilly" Transducer, Stopcock, Short No. 24 Needle, and 258 cm. No. 19 Polyvinyl Catheter.

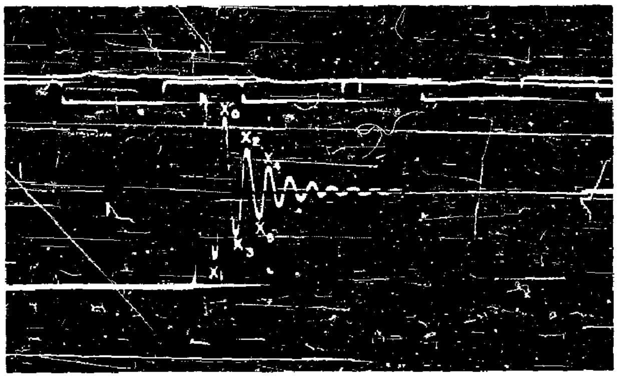


Figure 9 - Oscillogram of Pressure Step Function Response Showing Measurements Used for Determination of Response.

driving amplitude Y. If the values of K/m and R/m can be found, the relative amplitude versus frequency response may be computed since m is constant.

The following gives the equation governing the simple, lumped, series resonant circuit driven by a step function

$$m\frac{d^2x}{dt^2} + R\frac{dx}{dt} + KX = S \qquad (3)$$

Where S is the height of the step function and all other quantities are the same as in Equation (1), the solution to this equation is found to be

$$x(t) = \frac{S}{m}\,\frac{e^{-\frac{R}{2m}t}}{\omega_{nd}\sqrt{\frac{K}{m}}}\,\sin(\omega_{nd}t-\phi)$$

$$\tan\phi' = \frac{\omega_{nd}}{\frac{R}{2m}} \quad \text{and} \quad \omega_{nd} \equiv \sqrt{\frac{K}{m}-\left(\frac{R}{2m}\right)^2} \qquad (4)$$

The appearance of this solution (response) is a rapid amplitude change on which is superimposed a sine wave whose maximum value decreases with time (Figure 9).

Direct measurement of x is not practical. However, a recording of the transducer output voltage may be scaled and measured. Since this voltage is proportional to x, its record may be used interchangeably with x when the response relative to static pressure response is desired. Suppose x is measured at a maximum or minimum of the sine wave where the final steady value of pressure is taken as the origin, then

$$\begin{aligned} &\sin \quad (\omega_{nd} t_{\ell} - \phi') = 1 \\ &\text{and} \quad \omega_{nd} t_{\ell} - \phi' = \frac{(2\ell+1)\pi}{2} \\ &\text{or} \quad t_{\ell} = \frac{(2\ell+1)\pi}{2\omega_{nd}} + \frac{\phi'}{\omega_{nd}} \quad , \end{aligned} \tag{5}$$

similarly, at another maximum or minimum

$$\begin{aligned} &\omega_{nd} t_{m'} - \phi' = \frac{(2m'+1)\pi}{2} \\ &\text{or:} \\ &t_{m'} = \frac{(2m'+1)\pi}{2\omega_{nd}} + \frac{\phi'}{\omega_{nd}} \end{aligned} \tag{5}$$

where l and m' are the integral number of peaks (positive or negative) starting with the first as 0, the second 1, etc. Therefore, substitution of Equation (5) into Equation (4) and taking the ratio of $x_{\ell}/x_{m'}$ gives

$$\frac{X_{\ell}}{X_{m'}} = \frac{\frac{S}{m}\frac{1}{\omega_{nd}\sqrt{\frac{K}{m}}}\, e^{-\frac{R}{2m}\left(\frac{(2\ell+1)\pi}{2\omega_{nd}} + \frac{\phi'}{\omega_{nd}}\right)}(\pm 1)}{\frac{S}{m}\frac{1}{\omega_{nd}\sqrt{\frac{K}{m}}}\, e^{-\frac{R}{2m}\left(\frac{(2m'+1)\pi}{2\omega_{nd}} + \frac{\phi'}{\omega_{nd}}\right)}(\pm 1)}$$

or

$$\frac{X_{\ell}}{X_{m'}} = \pm\, e^{-\frac{R}{2m}\left(\frac{(2\ell+1)\pi-(2m'+1)\pi}{2\omega_{nd}}\right)} = \pm\, e^{-\frac{R}{2m}\frac{(m'-\ell)\pi}{\omega_{nd}}} \qquad (6)$$

Where m'-l is the number of intervals of adjacent maxima and minima between the occurrence of x_l and $x_{m'}$.

Now $$\mathrm{Ln}\left|\frac{X_{\ell}}{X_{m'}}\right| = \frac{R}{2m}\frac{(m'-\ell)\pi}{\omega_{nd}} \qquad (7)$$

Therefore, $$\frac{R}{m} = \frac{2\omega_{nd}\,\mathrm{Ln}\left|\frac{X_{\ell}}{X_m}\right|}{(m'-\ell)\pi}$$

Now, $\omega_{nd}, \frac{X_{\ell}}{X_m},$ and m'-l are all measurable from the step function response, so $\frac{R}{m}$ is calculable.

And since $$\omega_{nd} = \sqrt{\frac{K}{m} - \left(\frac{R}{2m}\right)^2} = 2\pi f_{nd}$$

Therefore, $$\frac{K}{m} = \omega_{nd}^2 + \left(\frac{R}{2m}\right)^2$$ is calculable. (8)

Substitution of Equations (7) and (8) in Equation (2) and variation of γ provide points on the desired amplitude versus frequency curve (relative to the response at static pressure, i.e., zero frequency). This method of obtaining the amplitude-frequency curve applies only when the system is "underdamped," i.e., when there are enough cycles of the natural frequency in the response to allow the measurement of the damped natural frequency $(\omega_{nd} = 2\pi f_{nd})$ and the decrement $\left(Ln\left|\frac{X_\ell}{X_m}\right|\right)$.

When the system is nearly "critically damped" or "overdamped," the response is not oscillatory, and the solution to the governing equation (for the overdamped case) is appropriately a hyperbolic function instead of a circular function. The amplitude-frequency curve may still be obtained from the step function response using the tables (Tables I and II) and method of Erik Warburg (24). Warburg has prepared a table of the time occurrence in the step function response, of the 40 per cent and 90 per cent of final amplitude points for various degrees of damping. Thus, the 40 per cent and 90 per cent amplitude can be located in time from the step function response and the ratio of these time values, compared to Warburg's Tables (24), to find R/m and

and K/m. (In Warburg's notation $\tau = \sqrt{\frac{K}{m}}\, t$, and $\alpha = \frac{R}{2\sqrt{mK}}$ α is first found in (24)). Table II by the ratio of $\frac{\tau_{A.4}}{\tau_{A.9}} = \frac{t_{A.4}}{t_{A.9}}$. Then τ_A is found in (24), Table II. Since t_A is known, $\sqrt{\frac{K}{m}}$ may be computed and multiplication of $\sqrt{\frac{K}{m}}$ by α gives $\frac{R}{2m}$. The record for the nearly critically damped system exhibits only a half oscillation (or less) of short duration which must be used for time measurement. Thus, for the nearly critically damped case, accurate time measurement is difficult.

C. DISCUSSION AND COMPARISON OF RESULTS FROM DIFFERENT TESTING SYSTEMS

Using step function responses and the above described methods, response curves were computed for pressure transducers of three basic types: (a) variable capacitance transducers (Lilly manometer and Sanborn electromanometer), (b) variable resistance transducers (Statham unbonded strain gauge), and (c) variable reluctance transducers (Wetterer-Gauer gauge). The transducer systems measured together with the various types and dimensions of catheters studied, and the response curves obtained for these various catheter-gauge configurations are included in Appendix B.

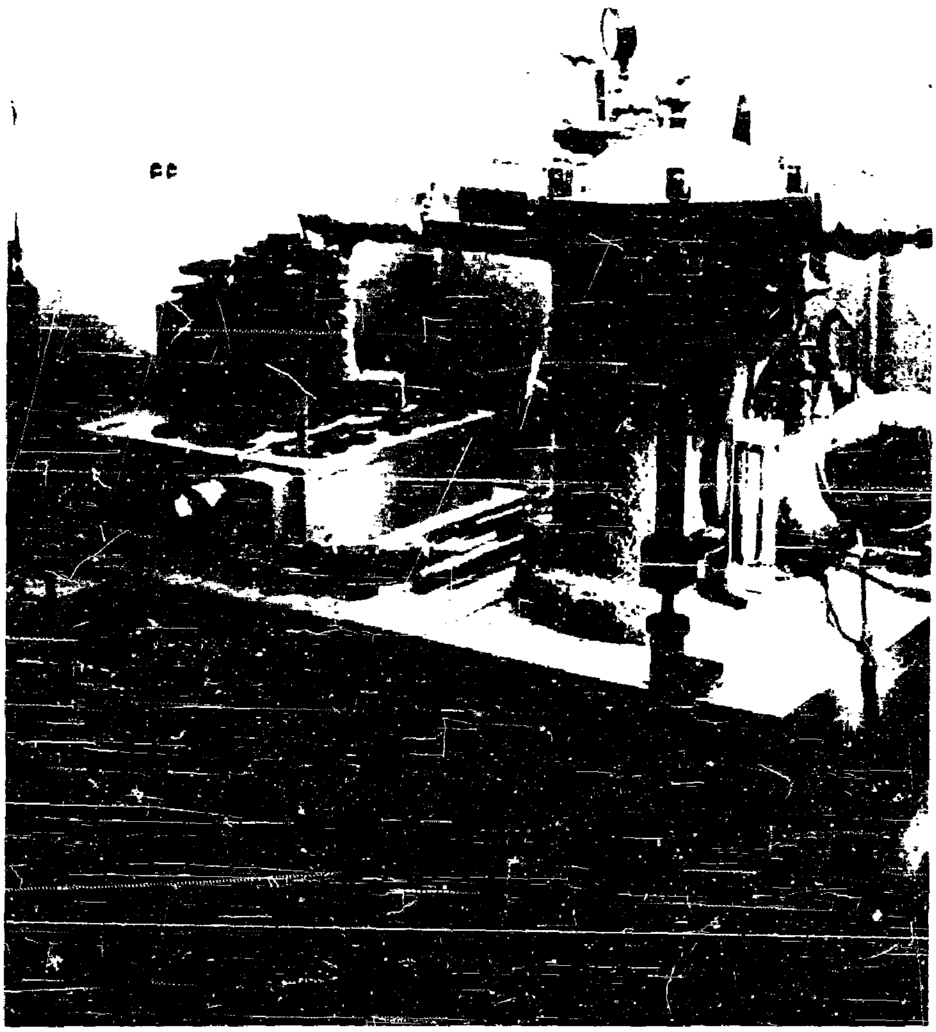


Figure 10 - Underwater Sound Reference Laboratory Low Frequency Calibration Tank.

It was necessary to validate the data obtained by the above methods. The Underwater Sound Reference Laboratory at Orlando, Florida, is operated by the U. S. Navy for the purpose of setting standards for all of their hydroacoustic equipment (sonar, etc.). The equipment used by this laboratory is ideal for checking the step function results. Consequently, amplitude-frequency responses of many combinations of transducers, fittings, and catheters were directly measured using the USRL facilities. There were two methods employed in these measurements. The free-field reciprocity method (20), carried out in a large lake, was used for the frequency range of 40 to 5000 cps. The low frequency tank (21) was used for the frequency range of 3 to 100 cps. (Figure 10). The transducers and combinations of transducers, fittings, and catheters studied by step function response methods were also calibrated by these absolute methods. Graphs of the results are included in Appendix B.

The "pistonphone" or pressure sine wave generator devices were also used for transducer evaluation. Two of these devices were available for test. The pistonphone (a small motor driven

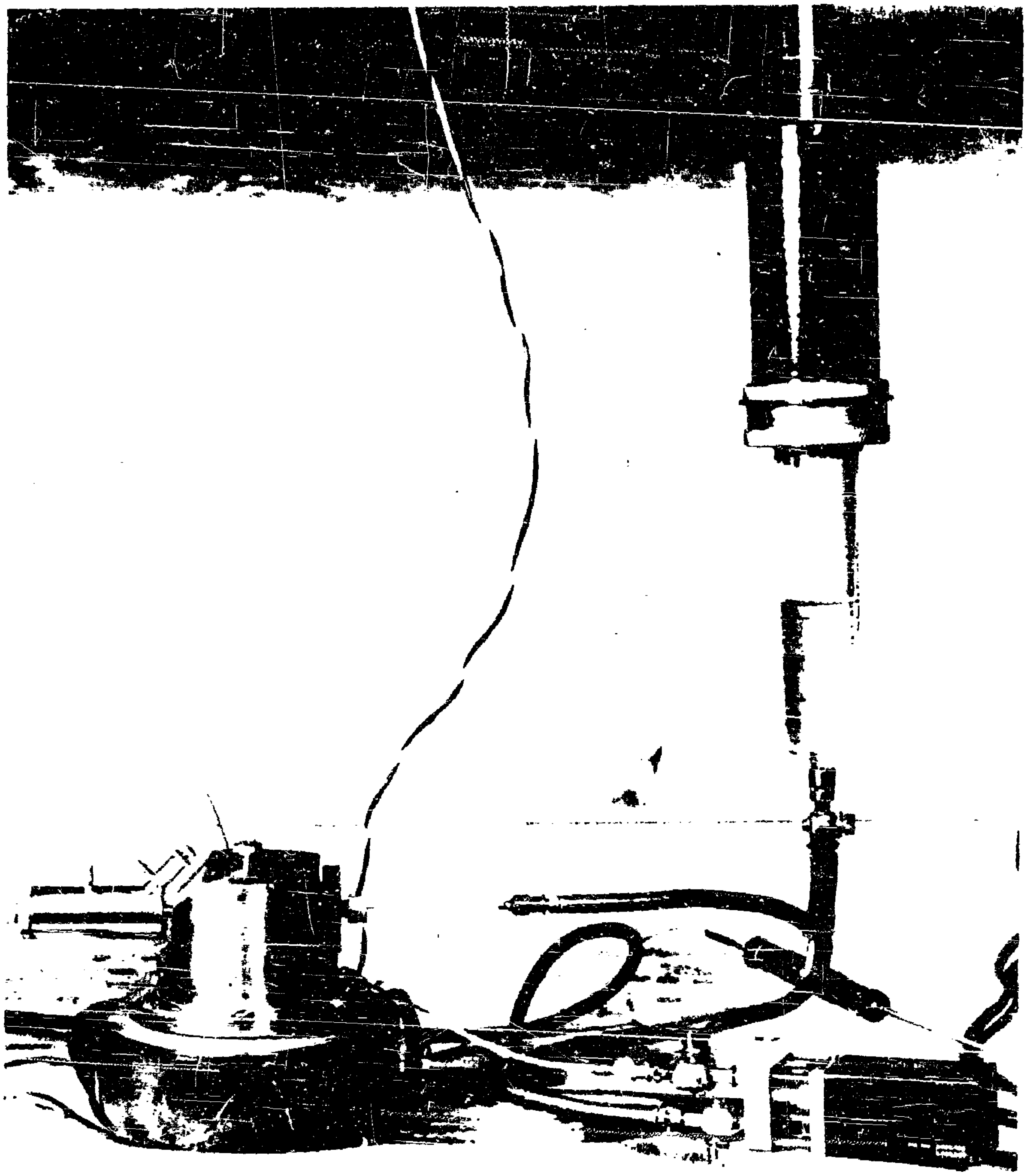


Figure 11 - AMAL Pistonphone Showing Monitoring Gauge (Lilly Capacitance Type) in Position and Statham Gauge Being Tested.

valveless pump), procured from the Basic Science Branch of the Mayo Foundation, was tested and found to impart too much mechanical vibration to the system under test. The Aviation Medical Acceleration Laboratory pistonphone was tested and thought to require some redesign along the lines of the Underwater Sound Reference Laboratory low frequency tank (21) before its use would be feasible. This redesign provided for a completely filled liquid chamber and for easier introduction of the manometer system into the pistonphone chamber. The rebuilt pistonphone was found to have a remarkably constant response from 7 to 700 cps. when one of the Underwater Sound Reference Laboratory tested transducers was used as a monitor (Figure 11). This pistonphone was then used to measure the amplitude-frequency responses of many of the transducer system combinations which were calibrated with the equipment at the Underwater Sound Reference Laboratory and with the step function method. Comparisons of the amplitude versus frequency response curves obtained by the step function pistonphone and absolute calibration techniques for the various transducers and combinations of transducers, fittings, and catheters have been made.

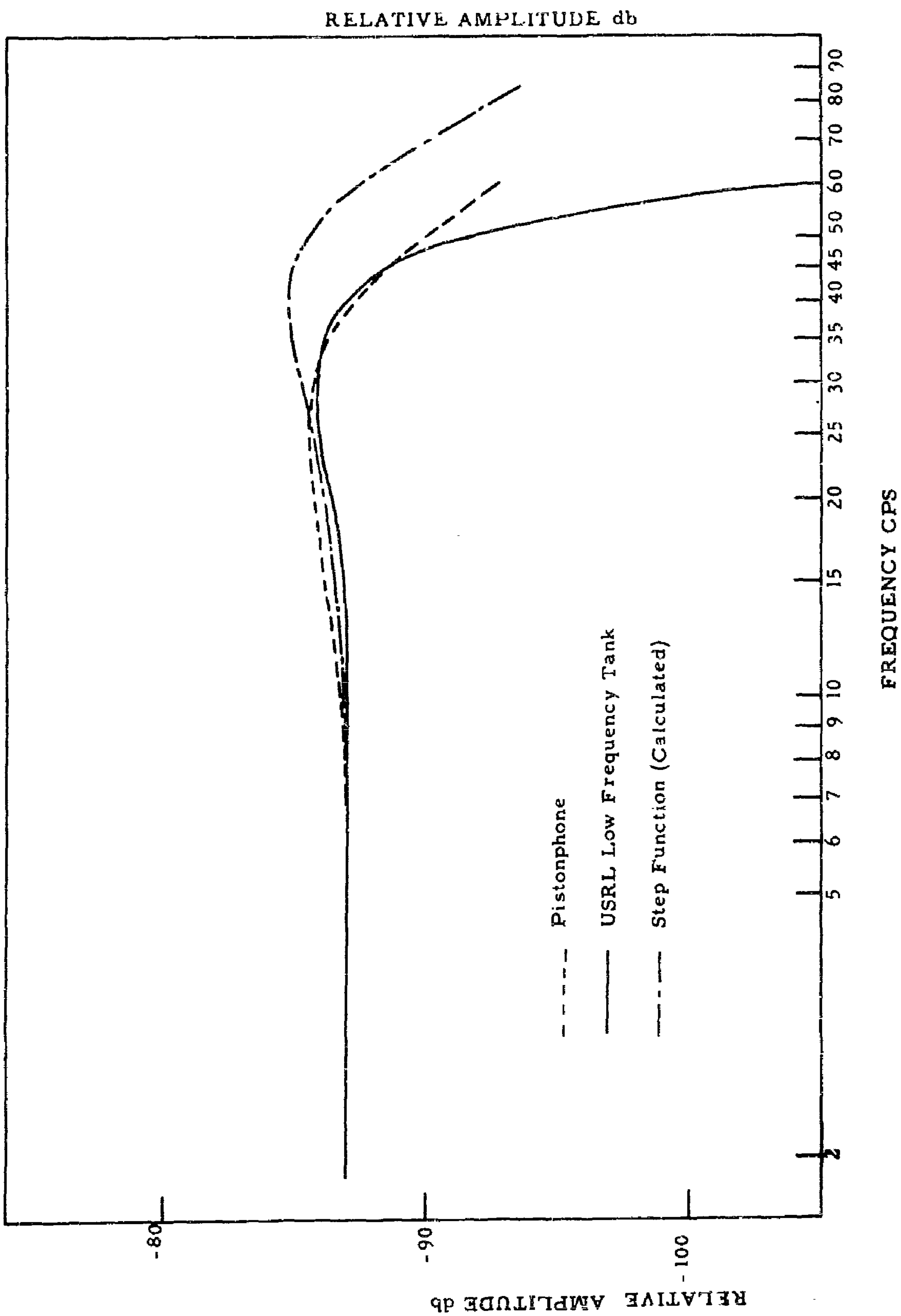


Figure 12 - Comparison of Three Methods of Response Measurement Using Sanborn Variable Capacitance Transducer.

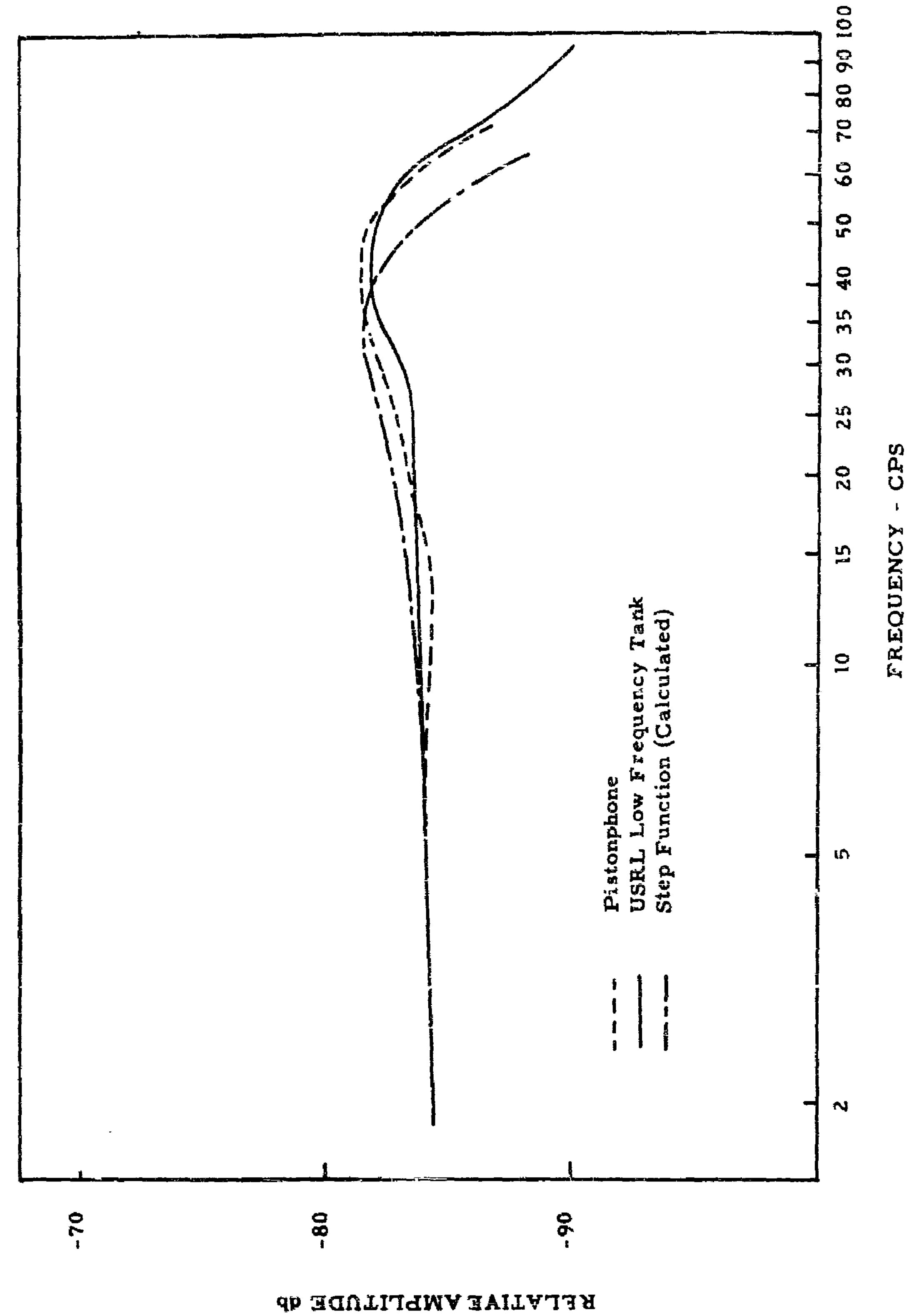


Figure 13 - Comparison of Three Methods of Response Measurement Using Lilly Variable Capacitance Transducer.

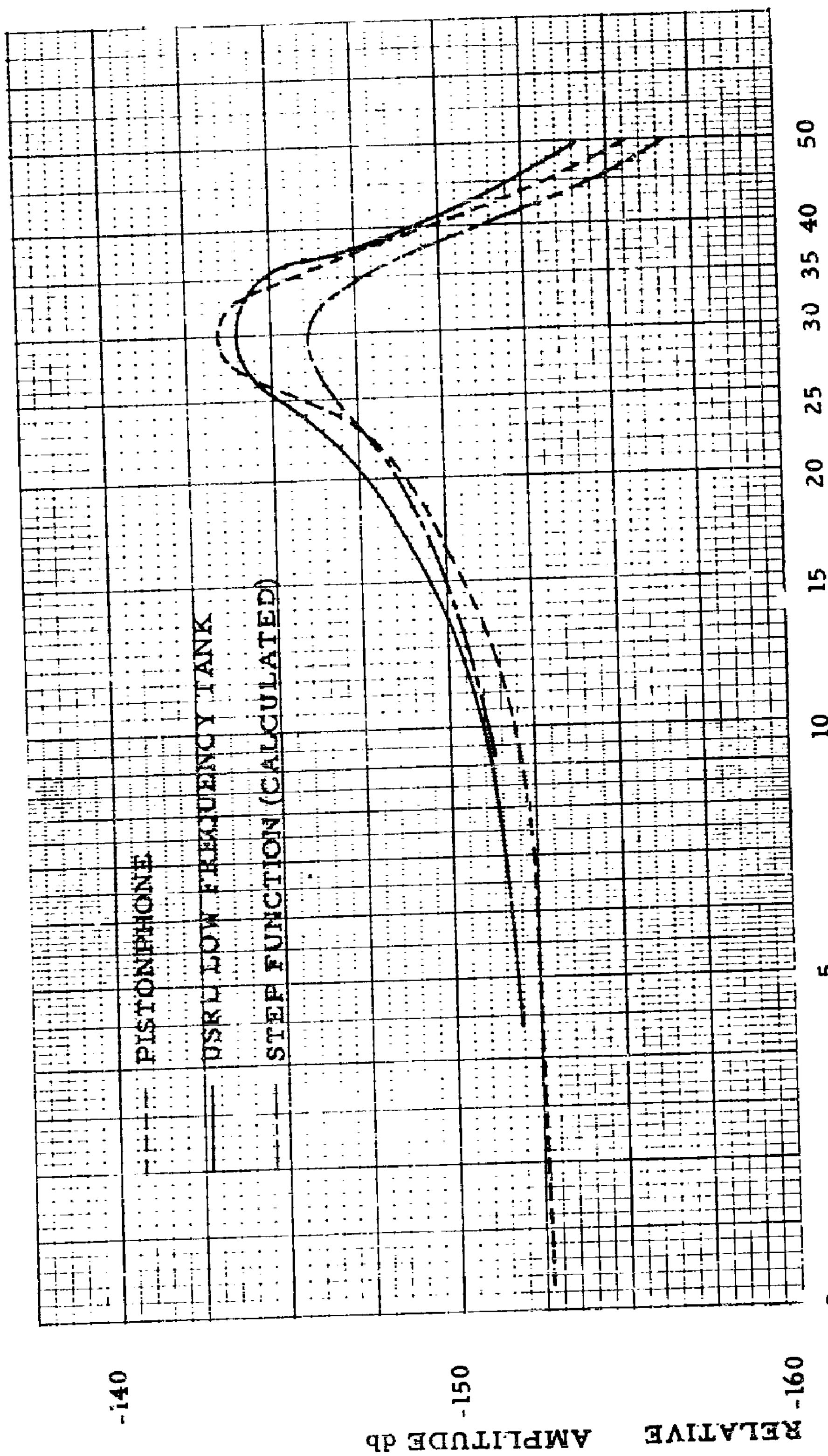


Figure 14 - Comparison of Three Methods of Response Measurement Using Statham Variable Resistance Transducer.

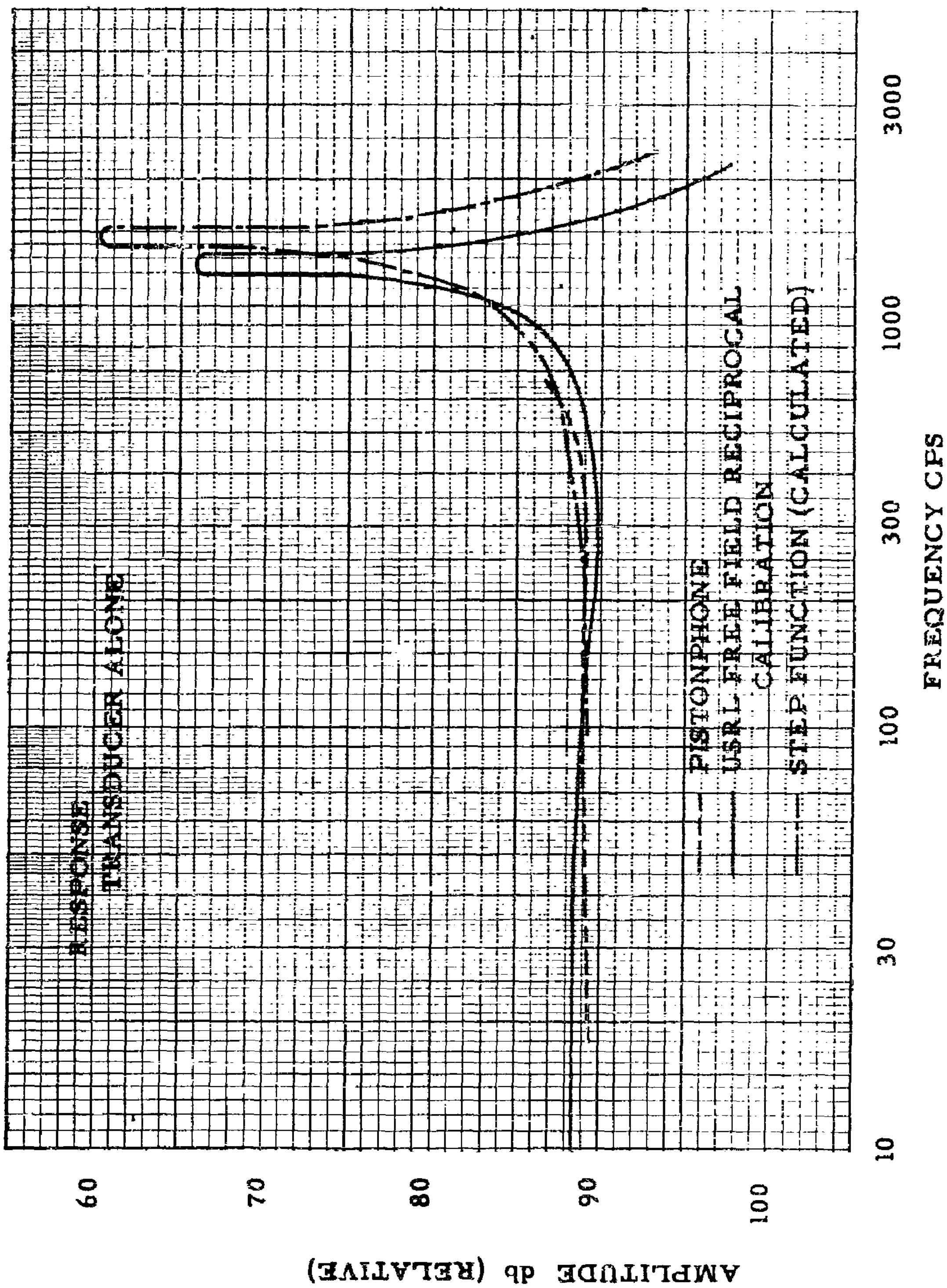


Figure 15 - Comparison of Three Methods of Response Measurement
Using Gauer Variable Reluctance Transducer.

A few of the results of the comparisons of the three methods are shown in Figures 12, 13, 14, and 15. Each figure is for a different type of manometer and the volume displacements of the manometers differ widely. Since the pistonphone and step function methods yield only relative responses, their output curves at and near the frequency of zero cycles per second were superimposed on the curves from the absolute method. Figures 12 and 13 show the response of a nearly critically damped system employing catheters. The disagreement of the responses calculated from step function records was thought to be due to the inability to accurately measure the quantities needed for the calculation in a nearly critically damped system. Figure 14 is a less damped system and shows better agreement, while Figure 15 shows the responses for a manometer which has a replaceable rubber diaphragm. The difference in resonant frequency is explained by differences in the rubber diaphragms. Also, the pistonphone response is shown only up to 700 cycles per second because the pistonphone monitor response ceased to be constant above that frequency. In nearly all cases, agreement of the three

methods of measurement was excellent. For example, the curves computed from step function responses of systems containing one meter of catheter showed good agreement with the Underwater Sound Reference Laboratory and pistonphone measurements suggesting that the system responded essentially as a one degree of freedom system.

From an analysis of these data, one concludes that, properly handled, the pistonphone and step function evaluation methods give a true description of the amplitude versus frequency response of a given manometer. Also, in view of the reliability of both methods, it is the opinion of these investigators that the observation of a manometer's response to a pressure step function is a better criterion for evaluation than an amplitude-frequency response, since all physiological pressure phenomena are non-sinusoidal in character.

The inference that the transducer systems studied could be considered for practical measurement purposes as a system having one degree of freedom was further tested by measurement of the velocity of propagation along the catheters. This measurement was accomplished by applying equivalent step functions to a short and a long catheter of equal interval dimensions simultaneously through a "Y" coupling. Both

catheters were connected to manometers which were in turn connected to a double beam oscilloscope. The records of the two systems were photographed, and the time delay between the appearance of this step function response through the two catheters, together with the difference in length of the catheters furnished data from which the velocity of propagation was calculated. It was found necessary to reverse the positions of the two manometers, make another record, and average the results of the two since the difference in time delay between separate manometers was enough to make a considerable error in the velocity. The velocity was found to be 980 meters per second. Therefore, the wave length at a frequency of 100 cycles per second was about 10 meters. Since the catheter was always one meter or less, it may be considered to be short in comparison to a quarter wave length, provided the frequencies of interest are all less than 100 cps. This fact also suggests that in transducer systems with catheters whose length is comparable to those measured, i.e. less than one meter the velocity of wave propagation is rapid enough so that the system may correctly be considered as essentially a lumped system.

In summary, standard methods of transducer system evaluation

were tested and compared with absolute methods. The standard methods were found to be entirely adequate and accurate. These findings further demonstrate that the theory of transducer response measurement based upon the assumption that the systems may be treated as simple, lumped systems.

PHASE II

DISTRIBUTED SYSTEMS

A. STATEMENT OF BASIC CONCEPTS

All of the preceding work was done with systems which were demonstrated to function as simple systems with one degree of freedom. In terms of the electrical analogs, they were considered as series resonant, lumped circuits. However, if catheters with a velocity of propagation of considerably less than 980 meters per second or of greater length than one meter are to be used, or if frequencies higher than 100 cycles per second are to be considered, then the catheter length becomes significantly long in comparison to a quarter wave length and the above assumptions for a lumped system may not be valid even for an approximation.

Therefore, it was thought desirable to provide a theory for the response of a transducer system consisting of a transducer and a catheter which would not be frequency limited. The solution to this problem was suggested from the use of electrical analogs for the transducer (6), and the application of electrical transmission line equations

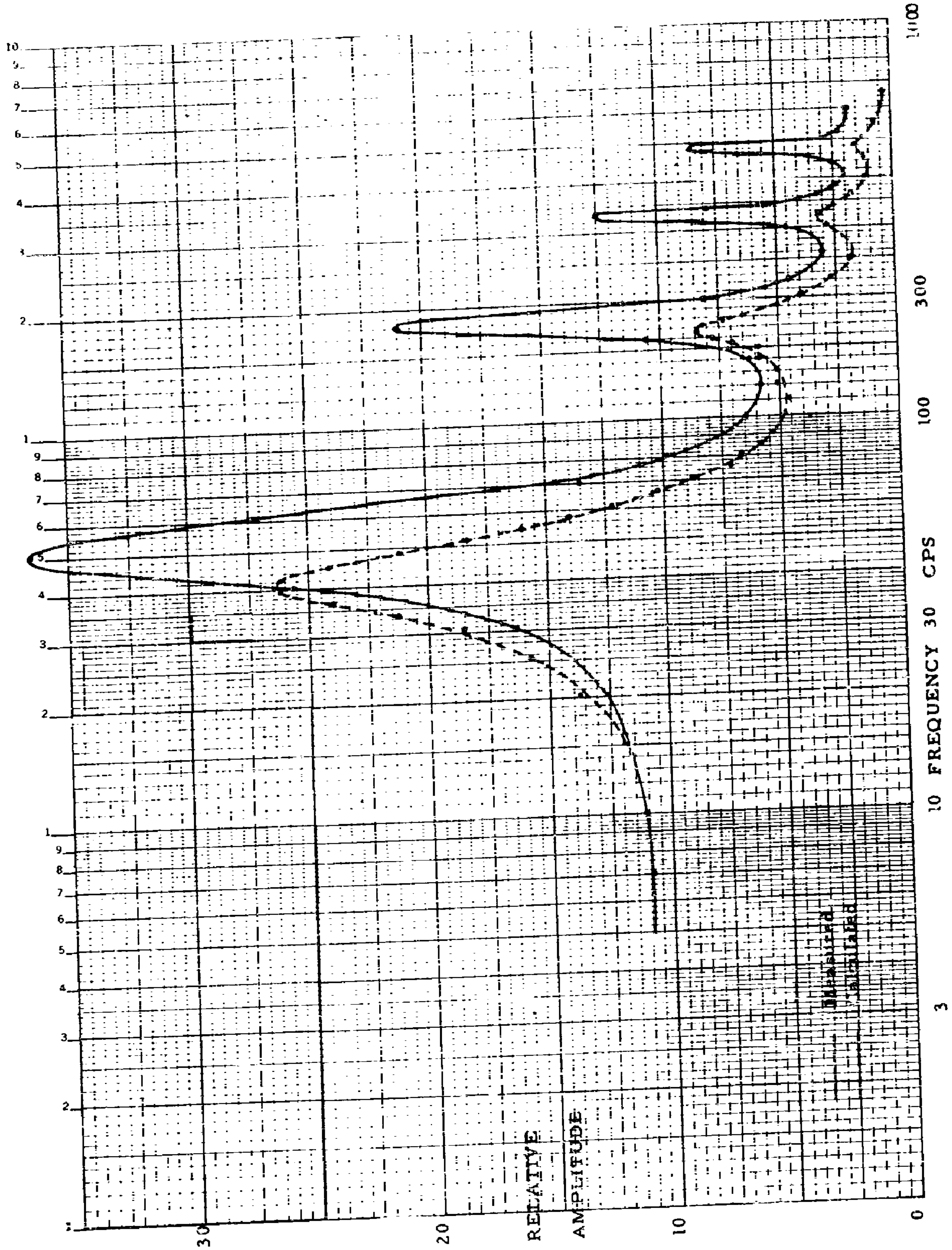


Figure 16 - Comparison of Calculated and Measured Amplitude Versus Frequency Response of Lilly Transducer, Stopcock, Needle Adapter, and 258 cm. Polyvinyl Plastic Catheter.

to the catheter.

The problem of calculating the amplitude-frequency response was approached by measuring the physical constants of the transducer (R, m, and K) and catheter (R_d, m_d, and K_d), and substituting them into the solution of the transmission line equations. The G_d of the transmission line was assumed to be zero since at static pressure, the velocity, or its analog (current) is zero because the catheter is made of nonporous material. If G_d was not zero, current could not be zero at static pressure if the analogs are valid.

The values of the other physical constants were then obtained, and the amplitude-frequency response of the system was calculated and compared to the amplitude-frequency response measured with the pistonphone. The results of this procedure for a 258 cm., 0.47 mm i.d., barium impregnated polyvinyl catheter No. 19 gauge used with a Technitrol "Lilly" transducer are shown on Figure 16. A more complete explanation of the procedure will be discussed later in this report.

The general differential equations for a transmission line without restricting its length or terminations are

$$L_d \frac{\delta i}{\delta t} + R_{ed} i = -\frac{\delta v}{\delta x}$$

$$C_d \frac{\delta v}{\delta t} + G_d v = -\frac{\delta i}{\delta x}$$

or in the analog of the transducer-catheter case

$$L_d \frac{\delta i}{\delta t} + R_{ed} i = -\frac{\delta v}{\delta x} \qquad (9)$$

$$C_d \frac{\delta v}{\delta t} = -\frac{\delta i}{\delta x} \quad ,$$

where R_{ed} and L_d are the resistance and inductance per unit length in series with the line and G_d and C_d are the conductance and capacitance per unit length across the line. The Laplace transform and transformation calculus are used to solve these differential equations (32 and 33). If there is no initial current flowing or initial charge on the capacitance per unit length, the line is said to be initially inert. The Laplace transformed equations for an initially inert line corresponding to equations (9) are

$$-\frac{\delta i}{\delta X} = pC_d \underline{v} \tag{10}$$

$$-\frac{\delta \underline{v}}{\delta X} = R_{ed}\underline{i} + pL_d\underline{i}$$

which combine to form the equation

$$\frac{\delta^2 \underline{i}}{\delta X^2} - \gamma^2 \underline{i} = 0 ,$$

where

$$\gamma = \sqrt{pC_d(pL_d + R_{ed})} \ . \tag{11}$$

If the source impedance is zero, the solution to this equation

$$\underline{i} = \frac{\underline{E}(p)}{Z_c} \frac{(Z_r + Z_c)e^{\gamma(\ell - X)} - (Z_r - Z_c)e^{-\gamma(\ell - X)}}{(Z_r + Z_c)e^{\gamma\ell} + (Z_r - Z_c)e^{-\gamma\ell}} \tag{12}$$

where E(p) = Laplace transform of the driving voltage

$Z_c = \sqrt{\frac{pL_d + R_d}{pC_d}}$

Z_r = the terminating impedance,

and ℓ = the length of the line.

Also,

$$\underline{v} = E(p)\frac{(Z_r + Z_c)e^{\gamma(\ell - X)} + (Z_r - Z_c)e^{-\gamma(\ell - X)}}{(Z_r + Z_c)e^{\gamma\ell} + (Z_r - Z_c)e^{-\gamma\ell}} \tag{12a}$$

In the case of the catheter-transducer system, the source impedance is effectively zero since the flow through the catheter is too small to affect the pressure at the point where the measurement is being made.

It is desired to find the amplitude versus frequency response curve when the free end of the catheter is excited by a variable frequency-pressure sine wave. This means that the amplitude of the volume of the transducer chamber is desired since the response of the system will be proportional to this quantity. The position amplitude of the liquid in the catheter at the transducer end multiplied by the catheter area is proportional to the transducer volume amplitude. The analog of the catheter position amplitude at the transducer end is the amplitude of the charge on the terminating capacitor (Figure 17). This charge amplitude is proportional to the voltage amplitude across the capacitor, since $v = \frac{q}{C}$. In this work, the analog of the voltage across the capacitor was not actually obtained. Instead, the analog of the current in the load was divided by the frequency (ω), which quantity is proportional to the analog of the voltage across the capacitor and, hence, is pro-

portional to the amplitude of the transducer chamber volume. The desired relation occurs at the transducer, so $x = l$, and remembering that for steady state sinusoidal excitation, $j\omega$ may be substituted for p, Equation (12) reduces to

$$\frac{I}{\omega E} = \frac{2}{\omega\left[(Z_r + Z_c)e^{\gamma\ell} + (Z_r - Z_c)e^{-\gamma\ell}\right]}; \tag{13}$$

Where E is the amplitude of the driving voltage and γ and Z become

$$\gamma = \sqrt{j\omega C_d R_{ed} - \omega^2 L_d C_d}$$

$$Z_c = \sqrt{\frac{L_d}{C_d} - j\frac{R_{ed}}{\omega C_d}} \tag{14}$$

The solution to the equation governing the mechanical system is the same as Equation (13). The equivalent circuit of the transducer-catheter system was assumed to be

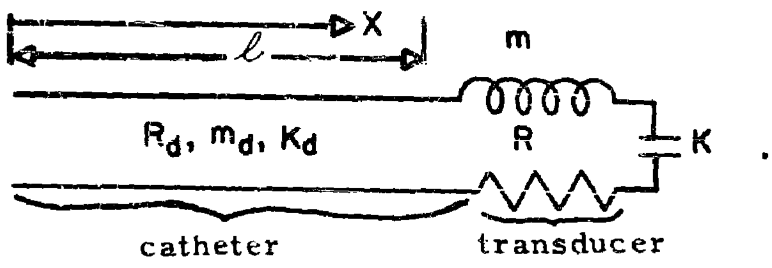


Figure 17 - Diagram of Electrical Analog of Transducer - Catheter System.

Z_r is recognized as the transducer which is a lumped system, and γ, Z_c, and Z_r become

$$\gamma = \sqrt{\frac{j\omega R_d - \omega^2 m_d}{K_d}} \;, \quad Z_c = \sqrt{m_d K_d - j\frac{R_d K_d}{\omega}} \;, \quad Z_r = R + j\omega m + \frac{K}{j\omega} \quad . (15)$$

B. MEASUREMENT OF PHYSICAL CONSTANTS OF SYSTEM

It was not considered practical to measure the physical constants of the line and transducer. R/m, ω_{nd} and K/m may be found using a step function as described in the section on the theory of lumped systems (pages 17 through 22). If either m or K can be found, all the values of R, m, and K may be calculated. If a variable capacitance type transducer is used as the instrument for test, the value of K can be determined by measuring the change in electrical capacitance for a given pressure change. The capacitance change, together with the total capacitance, and transducer geometry, allows the calculation of the volume change for the given pressure change (34). This provides the pressure to volume ratio which, when multiplied by the reciprocal of the square of the transducer opening area, is equal to K—the transducer spring constant. The Technitrol "Lilly" transducer was selected as the transducer to be studied.

The transducer was arranged with a glass capillary tube and a hypodermic syringe as indicated in the following drawing

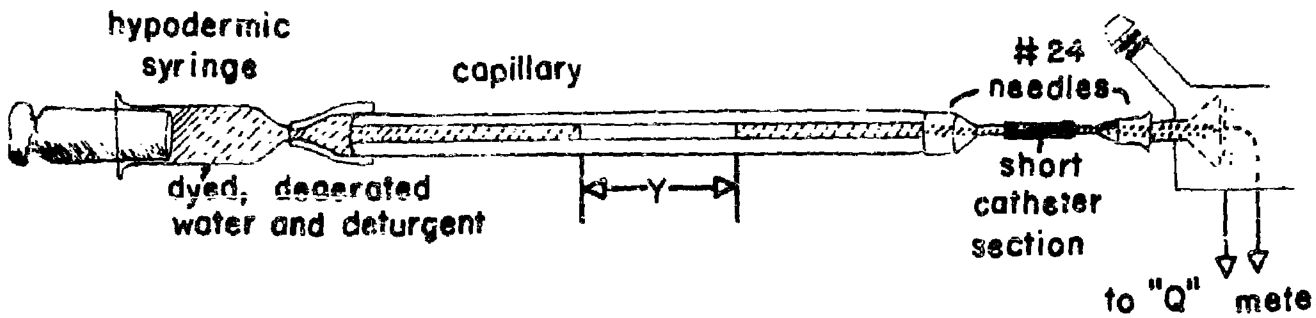


Figure 18 - Diagram of System Used to Measure Change of Capacitance with Change of Pressure in Variable Capacitance Transducer.

The initial capacitance of the transducer was measured on the Boonton "Q" meter. The syringe plunger was then pressed until the bubble in the capillary indicated by Y was reduced to half of its original volume indicating one atmosphere of pressure differential. The total capacitance was again measured and the capacitance difference noted. The actual measurements for one atmosphere were initial capacitance = 44.5 mmf, and capacitance difference = 4.0 mmf average.

Using the formula

$$C_o = \frac{\epsilon S}{4\pi D} \qquad \text{(Reference 34)} \qquad (16)$$

where ϵ = mks unrationalized specific inductive capacity

S = surface area $\pi r^2 = .1296 \times 10\pi$ Cm^2 ,

for the Technitrol transducer

D = air gap length,

the spacing D between the piates of the transducer capacitor was computed. $D = 8.11 \times 10^{-4}$ cm

From Lilly etal (26), the following formulae were obtained

$$S = \frac{C'}{C_o}\left(\frac{B}{A'}\right)^2 \tag{17}$$

$$= .0516 ,$$

where S = relative capacitance signal

C' = capacitance difference = 4.0 mmf

C_o = initial capacitance = 44.5 mmf

A' = radius of diaphragm = .475 cm, and

B = radius of fixed plate = .36 cm.

Also,

$$S = \frac{1}{\mu}\tanh^{-1}\left(\frac{\mu N}{1-WH}\right), \tag{18}$$

where $W = Y_o/D$

$W = .15$

$\mu = \sqrt{W}$

$N = (B/A')^2 = .575$

$H = 1 - N = .425$.

Y_o = deflection of the diaphragm at the center

w = Y_o/D (19)

Y_o = 1.218×10^{-4} cm for one atmosphere

vol = $\frac{1}{3}\pi A^2 Y_o$ for the stiff diaphragm (20)

$$= .262 \times 10^{-4}\ \mathrm{cm}^3/\mathrm{atmos.} = 3.45 \times 10^{-6}\ \mathrm{cm}^3/100\ \mathrm{mm\ Hg},$$

where vol = volume change for given deflection y_o.

The value of w was found from Equation (18) by choosing values of w and substituting in Equation (18) until an identity was found. Thus, the volume change for a pressure differential of 100 mm Hg. was found to be 3.45×10^{-6} cm^3/100 mm Hg. However, the volume change due to the flexing of the diaphragm is only a portion of the total volume change. This total is so small that the change in volume arising from the compressibility of water becomes an appreciable part. The volumes of the various parts of the transducer include

Chamber	.35 cm^3
Narrow part of stopcock	.079 cm^3 (d = .193 cm, l = 2.7 cm.)
Miscellaneous part of stopcock and nipple	.075 cm^3 (d = .416 cm, l = .3 + .25)

Narrow part of transducer appreciable	$.0297\ cm^3$ (d = .159 cm, l = 1.5 cm)
Total volume	$.534\ cm^3$.

The compressibility of water is 49.1×10^{-6} per atmosphere at 20°C. (35). Therefore, the volume change arising from the compressibility of water for a pressure differential of 100 mm Hg. is $3.45 \times 10^{-6}\ cm^3/100$ mm Hg.

The total volume change is, therefore

$$\frac{\Delta vol}{\Delta P} = 6.9 \times 10^{-6} cm^3/100\,mm\,Hg \quad \text{or} \quad 51.7 \times 10^{-12} cm^3/dyne/cm^2 \,. \qquad (21)$$

The ratio of the volume change to the pressure differential is found to be

$$\frac{\Delta vol}{\Delta P} = \frac{AX}{\frac{F}{A}} = A^2 \frac{X}{F} \,, \qquad (22)$$

where A is the cross sectional area of the portion of the transducer (the opening) where it is desired to refer the values of the analogs of R, m, and K. From Hook's law:

$$F = KX \,. \qquad (23)$$

Therefore, from equation (22)

$$\frac{1}{K} = \frac{\Delta vol}{\Delta P} \frac{1}{A^2} \,. \qquad (24)$$

It was now desired to obtain the effective mass of the transducer —an explanation follows, (27). The configuration of a variable capacitance transducer with a narrowed opening, as shown in the following Figure, was assumed.

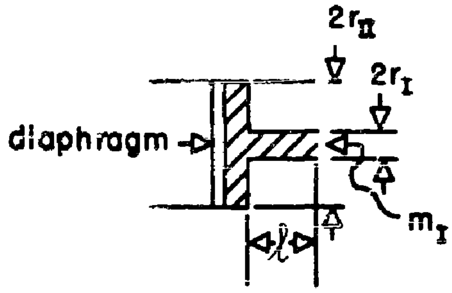


Figure 19 - Diagram of Variable Capacitance Transducer with Narrowed Opening.

Assuming the kinetic energy of the diaphragm and of the liquid immediately in front of it are negligible, then

$$K.E. = \frac{1}{2} m_I V_I^2$$

$$= \frac{1}{2} \pi r_I^2 \ell \rho V_I^2 ,$$

where V_I is the velocity of the mass m_I in the narrowed section, and ρ is the density of the liquid. If V_{II} is the velocity of the liquid in a transducer without a narrowed opening, then

$$V_I = \left(\frac{r_{II}}{r_I}\right)^2 V_{II} \tag{25}$$

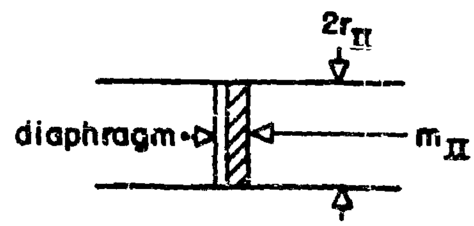


Figure 20 - Diagram of Variable Capacitance Transducer without Narrowed Opening.

since for the case being considered, the liquid may be considered as incompressible. Suppose the driving force amplitude, frequency, and spring constant for the systems represented by Figures 19 and 20 are kept constant, then, since the spring constant and driving force are the same, the peak potential energy of Figure 19 equals the peak potential energy of Figure 20. Therefore, if it is desired to make the systems of Figures 19 and 20 equivalent —the peak kinetic energies of the two must be equal.

$$\frac{1}{2} = m_I V_I^2 = \frac{1}{2} m_{II} V_{II}^2 \tag{26}$$

Substituting Equation (25) into Equation (26)

$$m_I\left(\frac{r_{II}}{r_I}\right)^4 = m_{II} \tag{27}$$

Thus, m_{II} must be $\left(\frac{r_{II}}{r_I}\right)^4$ times m_I for the two systems to be equivalent. It is seen that if the mass is being "referred" to a small diameter portion of the transducer (i.e., the opening of the transducer or the needle), the mass of a larger diameter portion must be multiplied by the fourth power of the ratio of the small to the large diameter before it may be added to the mass of the small diameter portion. The effective mass of the transducer referred to the chosen diameter is the sum of the masses of all portions of the transducer treated as described in the last sentence. The effective mass of the Technitrol Lilly transducer plus a three-way stopcock, and a .0308 cm i.d. by .95 cm long needle (No. 24 gauge) calculated from the above is 8.03×10^{-4} grams when referred to the inside diameter of the needle.

The value of $1/K$ of Equation (24) referred to the No. 24 gauge needle i.d. was found from the value of $\frac{\Delta vol}{\Delta P}$ of Equation (21) to be $93.2 \times 10^{-6} sec^2/gram$. Using this value and the natural frequency measured from the step function response (assuming no resistance) in the equation

$$\omega_n = 2\pi f_n = \sqrt{\frac{K}{m}} , \qquad (28)$$

the mass of the transducer (referred to the inside diameter of the No. 24 gauge needle) was found to be 10.7×10^{-4} grams. It is to be noted that the assumption of zero resistance is justified since the transducer-stopcock-needle system is very much under-damped and, therefore, the damped natural frequency differs inappreciably from the natural frequency (Figure 7). Comparison of this measured value of m with the calculated value above shows it to be 26 percent greater than the calculated value.

The value of $\frac{R}{m}$ for the Technitrol Lilly transducer-stopcock No. 24 gauge needle combination was found to be 515/sec. from the application of Equation (7) to measurements from the step function response of the combination. R was found by multiplying $\frac{R}{m}$ by the value of m calculated from the measured values of $\frac{I}{K}$ and ω Equation (28) was found to be .551 gram/sec. The values of R, m, and K of the transducer were now known. It was still necessary to get the values of R_d, m_d, and K_d of the catheter before the amplitude-frequency response of the system could be computed from the transmission line theory.

The change in volume for a given pressure change of the

catheter was measured in a manner similar to the method used to measure the same quantity for the transducer. The 258 cm No. 19 gauge catheter was assembled with the glass capillary tube and a hypodermic syringe filled with colored water, and plugged with a sewing needle. A bubble was introduced into the middle of the capillary making the entire assembly appear as follows

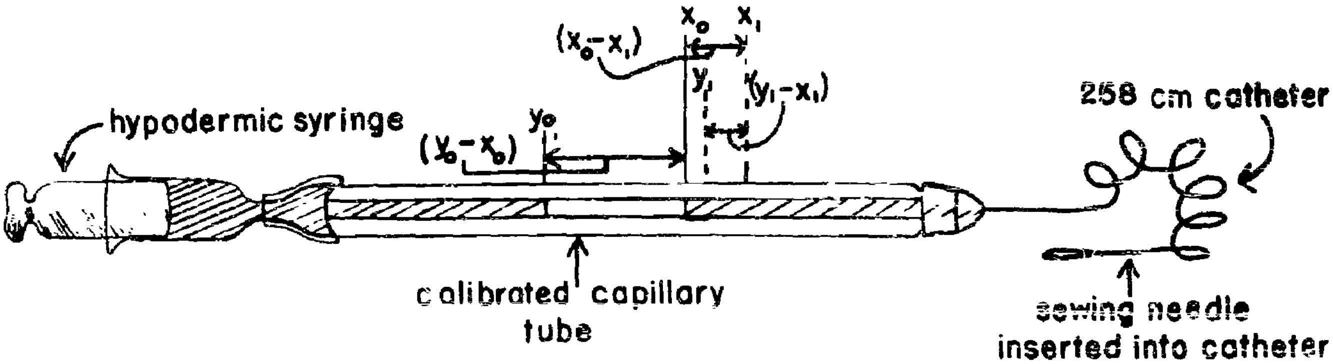


Figure 21 - Diagram of System Used to Measure Change of Volume of Catheter with Change of Pressure.

The syringe was pressed until the bubble was reduced to half of its original size ($Y_o - X_o = 2 (Y_1 - X_1)$, (Figure 21). However, both the size and the position (X_o and X_1) of the edge of the bubble near the catheter were observed. Assuming the assembly to be air free, and assuming that the assembly is rigid except for the catheter, the volume of the capillary between the

distal edge of the bubble before compression (X_0—Figure 21), and the distal edge of the bubble after compression (X_I—Figure 21) represents the volume change of the catheter for one atmosphere of pressure. It was confirmed that the assembly was air free to the left of the bubble by measuring the same volume change per atmosphere when the syringe plunger was withdrawn until the bubble was double size (one-half an atmosphere). Also, the assembly was checked for leaks and sticking by making sure that the bubble came back to the same size and position when the syringe had been released after compression or withdrawal. Only a very slight amount (less than 5 percent) of "creeping," Hansen (6), of the bubble due to volume change of the visco-elastic type in the catheter was observed.

The capillary diameter (as were all the small parts dimensions) was measured on a toolmakers microscope which had a laterally and transversely movable stage, the position of which could be measured to one ten-thousandth of an inch. This was convenient since the capillary bore turned out to be elliptical in cross section, and both axes had to be measured. On the basis of this measurement and the measurement of the positions of the

left edge of the bubble for a pressure differential of one atmosphere, the volume change of the 258 cm catheter was found to be 79×10^{-6} cm^3/atmosphere/258 cm catheter. By rearranging units

$$\frac{\Delta vol}{\Delta P} = .000079 \text{ cm}^3/\text{atmosphere}/258 \text{ cm catheter}$$

$$= .00003 \text{ cm}^3/\text{atmosphere}/100 \text{ cm catheter}$$

$$= .3 \times 10^{-6} \text{ cm}^3/\text{atmosphere}/\text{cm catheter}$$

$$= .296 \times 10^{-12} \text{ cm}^3/\text{dyne}/\text{cm}^2/\text{cm catheter} = .296 \times 10^{-12} \frac{\text{cm}^3 \text{ sec}^2}{\text{gram}} .$$

This $\frac{\Delta vol}{\Delta P}$ per cm is similar to $\frac{\Delta vol}{\Delta P}$ introduced earlier in (Equation (21)), and $1/K_d$ per cm can be calculated from Equation (24), where A is now the cross sectional area of the catheter. The area of the 258 cm catheter was calculated from its measured diameter, and was found to be .00173 cm^2. Thus, the value of $1/K_d$ per cm was

$$\frac{1}{K_d} = \frac{\Delta vol}{\Delta P} \frac{1}{A^2} = .0987 \times 10^{-6} \text{sec}^2/\text{g cm} . \tag{30}$$

The area calculated allows the calculation of an approximation of the mass per unit length m_d, which is .00173 gram/cm. The velocity of propagation along a lossless transmission line is $1/\sqrt{L_d C_d}$,

(29), m_d and $1/K_d$ are the analogs of L_d and C_d, respectively. The propagation velocity as calculated from this approximate mass was 765 meters per second, as compared to the measured velocity of 980 meters per second. A third value of velocity was obtained from the spacing of the peaks (excluding the first one) along the amplitude-frequency response curve of the 258 cm catheter as measured on the pistonphone. The response of the system indicated that the three peaks following the first corresponded to the first three resonance modes of the analogous transmission line shorted at both ends. The frequency spacing of the peaks and the length of the catheter allowed the calculation of the velocity from the equation

$$f\lambda = V, \qquad \text{where} \qquad (31)$$

(λ is twice the length of the catheter). The propagation as measured by this method was found to be 830 meters/sec. The velocities calculated from $1/\sqrt{L_d C_d}$ and measured from the amplitude-frequency curves are, as will be seen, for the higher frequency part of the spectrum. The velocity measured directly is for the low frequency part of the spectrum since only the very first

part of the step function response is considered, and for underdamped systems low frequencies are propagated more rapidly than higher ones. The mass per unit length for calculation purposes was computed from the velocity obtained from the amplitude-frequency curve, the measured value of K_d, and the equation

$$V = \sqrt{\frac{K_d}{m_d}} \tag{32}$$

This was done because it was felt that the mass obtained from either the velocity as measured, or by computation were not as accurate as that obtained from the amplitude-frequency curve velocity. The value of m_d computed from Equation (32) was .00147 gm/cm. The theoretical justification of this procedure will be considered in the section on calculation of distributed system response.

The resistance per unit length R_d which is the analog of the resistance per unit length of the electrical transmission line R_{ed} was calculated from the measured volume flow per unit time through a length of catheter for a given static pressure differential. The transmission line equation which contains R_d is

$$m_d \frac{\delta V}{\delta t} + R_d V = -\frac{\delta F}{\delta X} \qquad (33)$$

where m_d = mass, R_d = resistance, both per unit length, v = velocity of an interface, and F = force.

In the measurement of volume flow mentioned above

$$\frac{\delta V}{\delta t} = 0 \ . \qquad (34)$$

Therefore,

$$R_d V = -\frac{\delta F}{\delta X} \quad \text{or} \quad R_d \frac{\delta X}{\delta t} = -\frac{\delta F}{\delta X} \quad \text{and} \quad R_d = -\frac{\frac{\delta F}{\delta X}}{\frac{\delta X}{\delta t}} \ . \qquad (35)$$

Now,

$$F = PA \text{ and } \left.\frac{\delta F}{\delta X}\right)_{\text{static}} = \text{constant} = \frac{PA}{\ell} \qquad (36)$$

where A and l are the area and length of the catheter lumen and P = pressure in dynes per cm^2.

Also,

$$X = \frac{vol}{A} \text{ and } \left.\frac{\delta x}{\delta t}\right)_{static} = constant = \frac{vol}{tA} \text{ or } Rd = -\frac{\frac{PA}{\ell}}{\frac{vol}{tA}} \ , \qquad (37),$$

where vol is the volume of flow, and t is the time in seconds, P, A, l, t, and volume may all be obtained from the measurement

mentioned above, so R may be computed. Actually, two lengths of catheters were used. One catheter was 258 cm long No. 19 gauge and the other was 1.8 cm long No. 19 gauge--both of the same original piece of tubing. Before the measurements were made, it was desired to find the most suitable value of pressure to force the liquid through the catheter. If the resistance of the catheter were constant as the velocity was varied, it would not matter what pressure was used. That the resistance was constant, however, could not be tactitly assumed. Therefore, it was decided, to calculate a maximum velocity from the known constants of the system and the measured amplitude-frequency response, (Figure 16) and to put enough pressure on the catheter to make the velocity equal to this maximum. By halving and quartering this pressure and at the same time measuring the flow volume, the degree of constancy can be found. The volume displacement of the transducer for 100 mm Hg pressure differential is 6.9×10^{-6} cm^3/100 mm Hg; that of the 258 cm catheter is 10.4×10^{-6} cm^3/258 cm catheter/100 mm Hg making the total volume displacement 17.3×10^{-6} cm^3/100 mm Hg. The area of the lumen of the catheter is A =

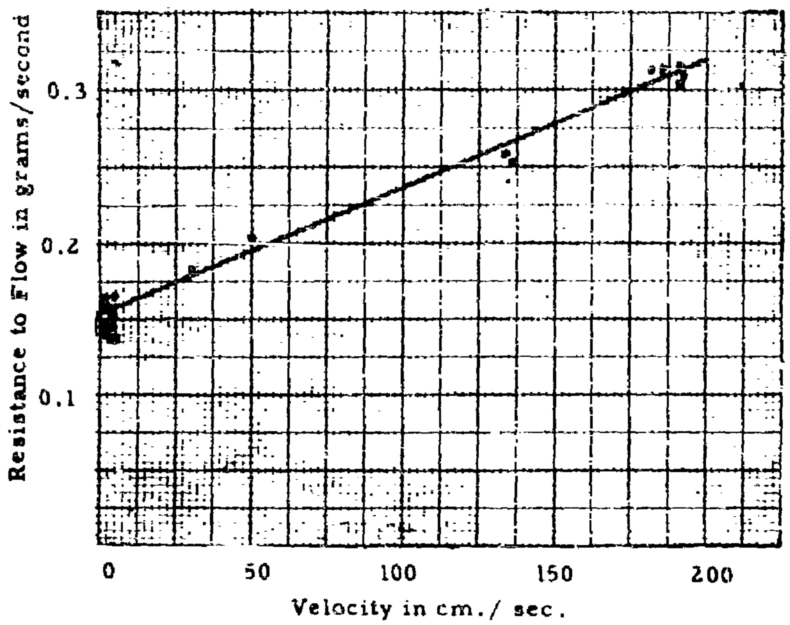


Figure 22 - Graph of Resistance to Flow Versus Interface Velocity 0.47 mm. ID Polyvinyl Catheter.

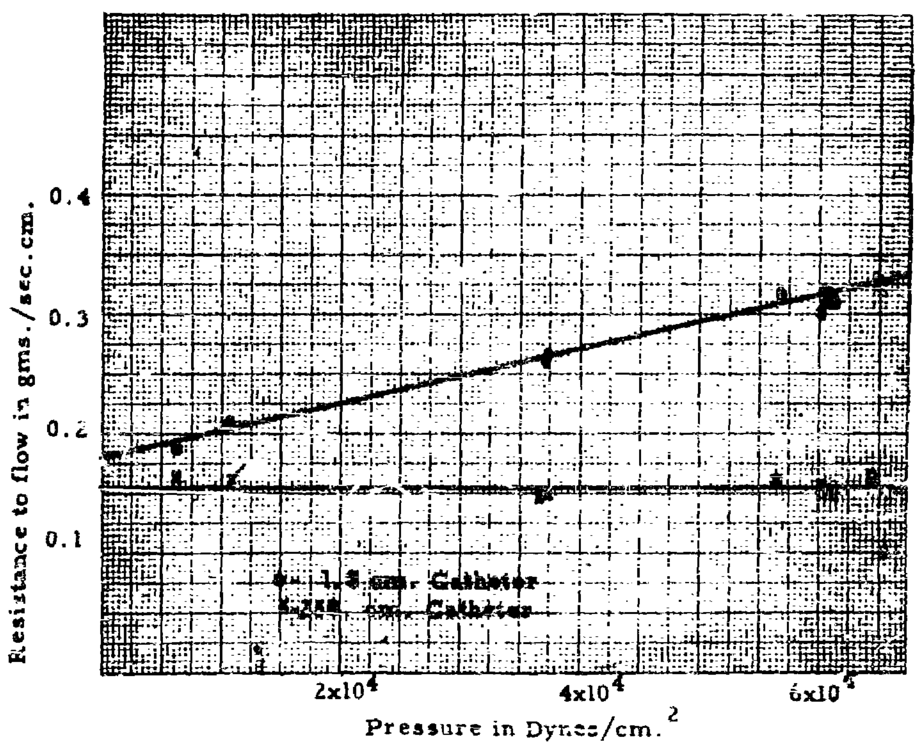


Figure 23 - Graph of Resistance to Flow Versus Applied Pressure in Two Lengths of 0.47 mm. ID Polyvinyl Catheter.

.00173 cm^2. Thus, the cross section of the liquid at the open end of the catheter moves a distance of 10^{-2} cm for 100 mm Hg static pressure. At very low frequencies, the position of an interface in the liquid is

$$X = 10^{-2} \sin \omega t \tag{38}$$

When this is differentiated, one finds

$$V = \frac{dx}{dt} = 10^{-2}\, \omega \cos \omega t \tag{39}$$

The first resonant frequency for this combination occurs at 41 cps. Figure 16, and the resonant peak rises above the static and low frequency pressure amplitude by a factor of 2.44. Therefore, the maximum velocity for this resonance is

$$V = 10^{-2} \times 82\pi \times 2.44 = 6.3\ \text{cm/sec} \tag{40}$$

which corresponds to a flow of

$$VA = 6.3 \times .00173 = .0109\ \text{cm}^3\text{/sec} = .65\ \text{cm}^3\text{/min}. \tag{41}$$

Figure 22 shows a plot of the resistance versus the interface velocity in the catheter, both quantities having been computed from the measurements of volume flow through the two lengths of catheter for various differential pressure values. It is seen that the resistance is constant for the maximum velocity of the first

resonance, although, it may be somewhat greater (approximately 15 percent) for the higher frequency resonances. The value of the resistance chosen for this work was .153 gram/sec. cm, which is seen to equal the resistance corresponding to the intercept of the curve of Figure 23 with the zero velocity axis. The reason for selecting this value is better seen from Figure 23 showing the resistance as a function of the applied pressure for the two lengths of catheter.

At this point, the values of m, K, and R referred to the needle diameter were known and the values of m_d, K_d and R_d referred to the catheter diameter were known. It was now necessary to transform the transducer values, (m, K, and R) to the catheter diameter, and then calculate the amplitude-frequency response from Equations (13) and(14). The transformation could be made by using the fourth power of the catheter to needle diameter ratio as described. (Equation 28.) It is easier, however, to use the electrical analog of an abrupt change in diameters which is an ideal transformer with an impedance ratio (transforming from the small to the large diameter) of the square of the large to small area ratio (29). This ratio, however, is the same as the fourth power of the large to small diameter ratio. The needle diameter is .308 mm, while the catheter diameter is .470 mm. Therefore, the fourth power of the ratio of the diameters is 5.44. Using this transformation ratio, the values of m, 1/K, and R referred to the catheter diameter are

$$\begin{aligned} m &= 58.1 \times 10^{-4} \text{ gram} \\ 1/K &= 17.15 \times 10^{-6} \text{ sec}^2/\text{gram} \qquad (42) \\ R &= 3.00 \text{ gram/sec} \;. \end{aligned}$$

In summary, the values of m_d, $^1/K_d$, and R_d referred to the catheter diameter are

$$
\begin{aligned}
m_d &= .00147 \text{ gram}/\text{cm} \\
1/K_d &= .0987\text{X}10^{-6} \text{ sec}^2/\text{gram cm} \qquad (43) \\
R_d &= .153 \text{ gram}/\text{sec cm} \quad .
\end{aligned}
$$

C. CALCULATION OF DISTRIBUTED SYSTEM RESPONSE

Equation (13) was rearranged in order to be able to use the "U. H. F. Transmission Line Chart" (30) in the computation of the amplitude-frequency response. After rearrangement, Equation (13) appeared as follows

$$\frac{I}{\omega E} = \frac{2}{\omega(Z_r - Z_c)} \; \frac{\dfrac{Z_r/Z_c - 1}{Z_r/Z_c + 1} e^{-\gamma\ell}}{1 + \dfrac{Z_r/Z_c - 1}{Z_r/Z_c + 1} e^{-2\gamma\ell}} \; . \tag{44}$$

Besides this, γ was transformed to

$$\gamma = \sqrt{\frac{j\omega R_d - \omega^2 m_d}{K_d}} = \sqrt{j\omega \frac{R_d}{K_d} - \omega^2 \frac{m_d}{K_d}} \tag{45}$$

$$= \left(\omega \sqrt{\omega^2 \left(\frac{m_d}{K_d}\right)^2 + \left(\frac{R_d}{K_d}\right)^2} \; e^{\; j\left(\pi - \tan^{-1} \frac{R_d}{\omega m_d}\right)}\right)^{\frac{1}{2}}$$

$$= \frac{\omega}{K_d} \left(\left\{m_d K_d\right\}^2 + \left\{\frac{R_d K_d}{\omega}\right\}^2\right)^{\frac{1}{4}} e^{\; j\left(\frac{\pi}{2} - \frac{1}{2} \tan^{-1} \frac{R_d}{\omega m_d}\right)} \; .$$

The computation of a point on the amplitude-frequency response curve was carried out as follows

1. The values of $|\gamma \ell|$ and $\angle \gamma \ell$ were found.

2. The values of $|Z_r|$ and $\angle Z_r$ were found.

3. Then, $\frac{Z_r}{Z_c}$ (complex) was found. ($|Z_c|$ and $\angle Z_c$ are by-products in the computation of $|\gamma \ell|$ and $\angle \gamma \ell$.)

4. And, from the "U. H. F. Transmission Line Chart," $\frac{Z_r/Z_c - 1}{Z_r/Z_c + 1}$ was found.

5. It was then necessary to find the values of $\frac{Z_r/Z_c - 1}{Z_r/Z_c + 1} e^{-\gamma \ell}$ and $\frac{Z_r/Z_c - 1}{Z_r/Z_c + 1} e^{-2\gamma \ell}$.

6. Then, $1 + \frac{Z_r/Z_c - 1}{Z_r/Z_c + 1} e^{-\gamma \ell}$ and its magnitude and angle were computed.

7. Finally, $|Z_r - Z_c|$ and $\angle Z_r - Z_c$ were found.

8. The above were then combined in the form

$$\frac{2\,\dfrac{Z_r/Z_c-1}{Z_r/Z_c+1}\,e^{-\gamma \ell}}{\omega(Z_r-Z_c)\left(1+\dfrac{Z_r/Z_c-1}{Z_r/Z_c+1}\,e^{-2\gamma \ell}\right)},$$

the magnitude of which is a point on the desired amplitude-frequency response curve.

This work was systematized in a Table which had 34 columns starting with the frequency and ending with the magnitude of 8. above. Each point on a curve required about 52 steps for its computation. As mentioned before, Figure 16 is a comparison of the calculated and the measured amplitude-frequency response of a Technitrol "Lilly" transducer, stopcock, No. 24 gauge needle, and 258 cms of No. 19 gauge (.47 mm ID) catheter. The discrepancies of the calculated curve when compared to the measured curve are seen to be a slightly higher frequency for the first resonance and considerably higher amplitude for all of the resonances. There are a number of possible reasons for these discrepancies, and they will be discussed in the section on the variation of response for changes in physical constants, pages 75 to 89.

Before continuing with the variation of the constants, however, it is well to justify the use of the resonant peaks of the pistophone amplitude-frequency response curve in the calculation of the catheter propagation velocity. Also, it is desirable to justify the use of $V = \sqrt{\frac{K_d}{m_d}}$ for the calculation of the mass per unit length of the liquid filled catheter. From the calculations of the points on the response curve, the following data have been determined:

ω	Z_r	Z_c
$2\pi 50$	$184 \angle -89.1^\circ$	$125.2 \angle -9.2^\circ$
$2\pi 180$	$48.05 \angle -86.2^\circ$	$121 \angle -2.6^\circ$
$2\pi 330$	$16.35 \angle -79.4^\circ$	$121 \angle -1.4^\circ$
$2\pi 480$	$3.86 \angle -34.9^\circ$	$121 \angle -1.0^\circ$

where the frequencies are all at the very peaks of the resonances. It is well known that if a transmission line is terminated in an impedance which is small compared to its characteristic impedance, it will act approximately as a shorted line. Since the driving point impedance is approximately zero, it is seen that the catheter will behave almost as a line shorted at both ends for frequencies above about 180 cps. This means that it will resonate when it becomes any multiple of a half wave length. This was the assumption made

in the calculation of the velocity of propagation. Also, from the calculations of the points on the response curve, the following data have been determined:

ω	$\left(\frac{R_d K_d}{\omega}\right)^2$	$(m_d K_d)^2$	$\angle\gamma$
$2\pi 50$	$.241\times10^8$	2.22×10^8	80.9°
$2\pi 180$	$.0186\times10^8$	2.22×10^8	87.4°
$2\pi 330$	$.006\times10^8$	2.22×10^8	88.6°
$2\pi 480$	$.003\times10^8$	2.22×10^8	89.0° .

Therefore, $|\gamma|$ is found to be

$$|\gamma| = \frac{\omega}{K_d}\left((m_d K_d)^2 + \left(\frac{R_d K_d}{\omega}\right)^2\right)^{\frac{1}{4}} \approx \frac{\omega\sqrt{m_d K_d}}{K_d} \approx \omega\sqrt{\frac{m_d}{K_d}} \approx \beta \tag{46}$$

for frequencies above about 50 cps. But for a "lossless" line (28)

$$\text{velocity} = \frac{2\pi f}{\beta} = \frac{\omega}{\beta} = \sqrt{\frac{K_d}{m_d}} = V\,. \tag{47}$$

This justifies the use of $V = \sqrt{\frac{K_d}{m_d}}$ for the calculation of the mass per unit length.

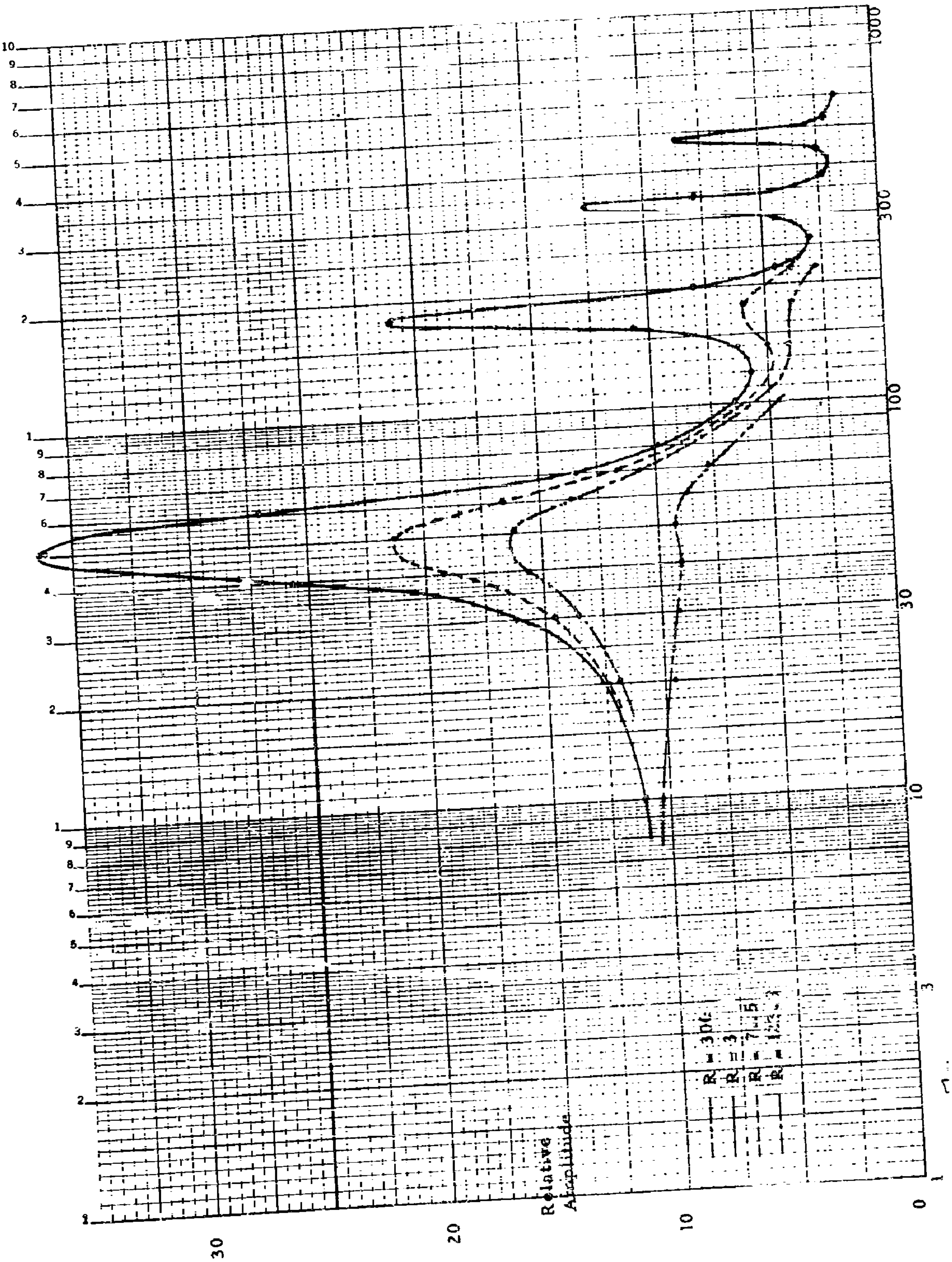


Figure 24 - Calculated Response of Transducer, Fittings, and Catheter as Resistance of Transducer R is Varied.

D. CALCULATION OF EFFECT OF VARYING PHYSICAL CONSTANTS

The general agreement in the form of the experimental and computed curves of Figure 16 justifies the empirical approach to the problem, and computation of more responses after arbitrary variation of one or more of the system constants might provide useful information of practical importance to those who require the ultimate in transducer response for their research methods.

In the simpler lumped systems without a catheter, it is an accepted technique to increase the resistance loss in the transducer in order to prevent the large response to frequencies near the resonance frequency. The usual procedure is to increase R until the system is nearly "critically damped" or until the amplitude does not increase with frequency above zero cycles per second (6). For this reason, it appeared logical to calculate the response of the distributed system assuming increased values of transducer resistance, R. Figure 24 shows theoretical curves for varying amounts of R. The top curve is the same as the theoretical curve

of Figure 16, i.e., the system without any increase in the value of R. It will be noted that a curve of the system with R about 40 instead of 3 would closely match the experimental amplitude-frequency curve of Figure 16. It is thought, however, that an error of this magnitude could not have been made in the experimental determination of R. Also, it is to be noted that with R equal to 300, the transducer itself is considerably overdamped, and its response superimposed on what would be a peaked response of the catheter, gives a curve which is almost flat but has the "double peaked" appearance found by Hansen (6).

Another means of preventing the amplitude from rising greatly at the first resonance is to increase the volume change of the transducer for a given pressure differential. This would correspond to decreasing the value of K. Since the velocity of the liquid in the catheter would be increased, the loss in the catheter should also be increased. Therefore, the response amplitudes at the resonant frequencies should be decreased. Of course, from Equation (4), a decrease in the value of K means a decrease in the frequency of the first resonance. Figure 25 shows theoretical

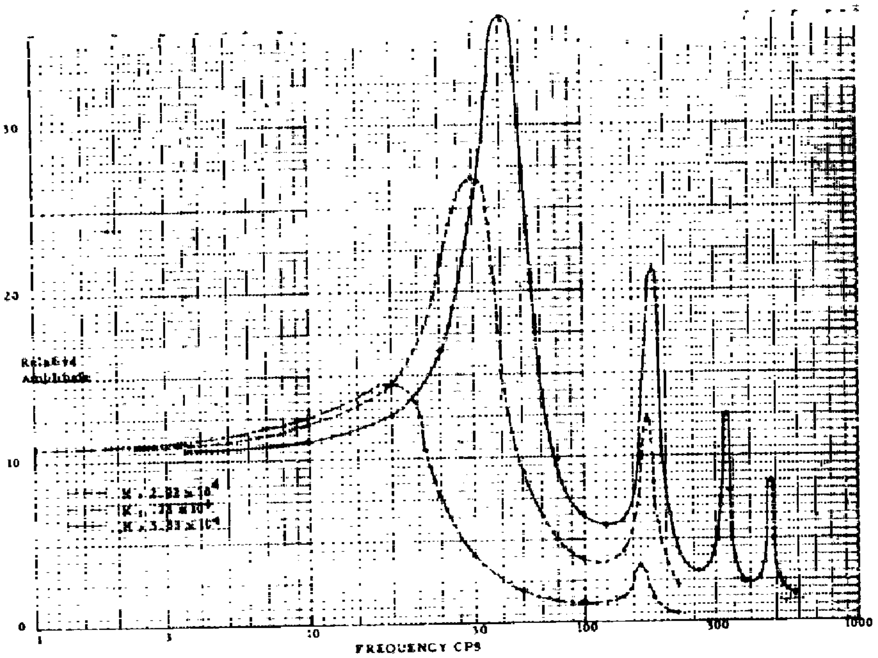


Figure 25 - Calculated Response of Transducer, Fittings, and Catheter as Spring Constant of Transducer K is Varied.

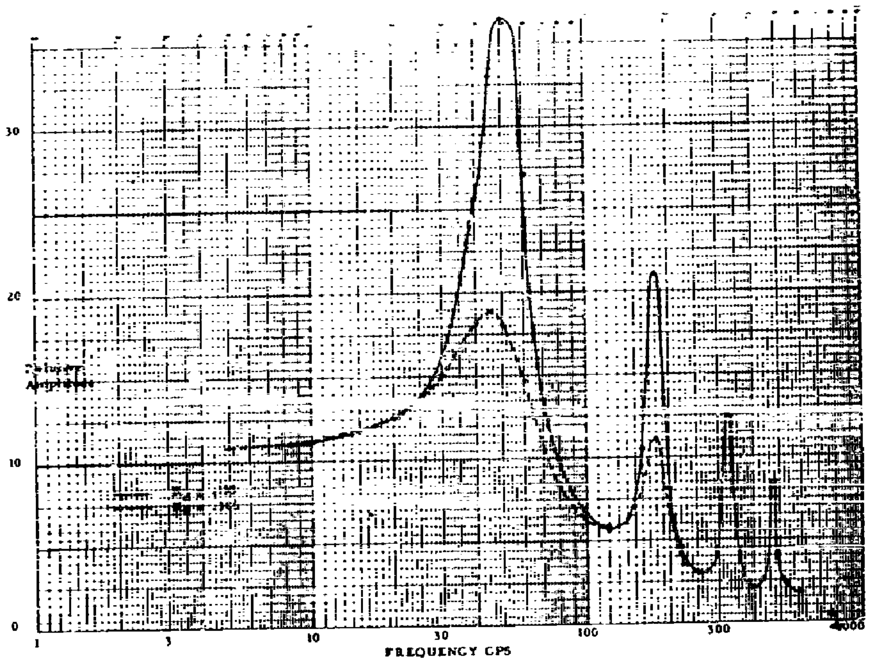


Figure 26 - Calculated Response of Transducer, Fittings, and Catheter as Resistance Per Unit Length of Catheter R_d is Varied.

amplitude-frequency response for the original system except for decreasing the values of K. The top curve (as it is in Figures 24, 25, 26, 27, 28 and 29) is the same as the theoretical curve of Figure 16

The next change to be made in the constants of the system is shown in Figure 26 where the theoretical amplitude-frequency responses for the normal and an increased value of R_d are plotted. The lower curve for the increased value of R_d represents the response of a system with the same geometry and material as the system of the upper curve, but in which a liquid with higher viscosity is used. If this curve is compared with the measured curve of Figure 16, it is seen that if the actual value of R_d was higher than the measured value, the theoretical curve would more nearly match the measured curve. However, in this case, a small error in the measured value of R_d makes a large difference in the theoretical curve, so it is thought that such an error may be an explanation of the discrepancies mentioned with reference to Figure 16.

Figure 27 shows the calculated effect of lowering the value of the spring constant per unit length, K_d. The curve with the lower

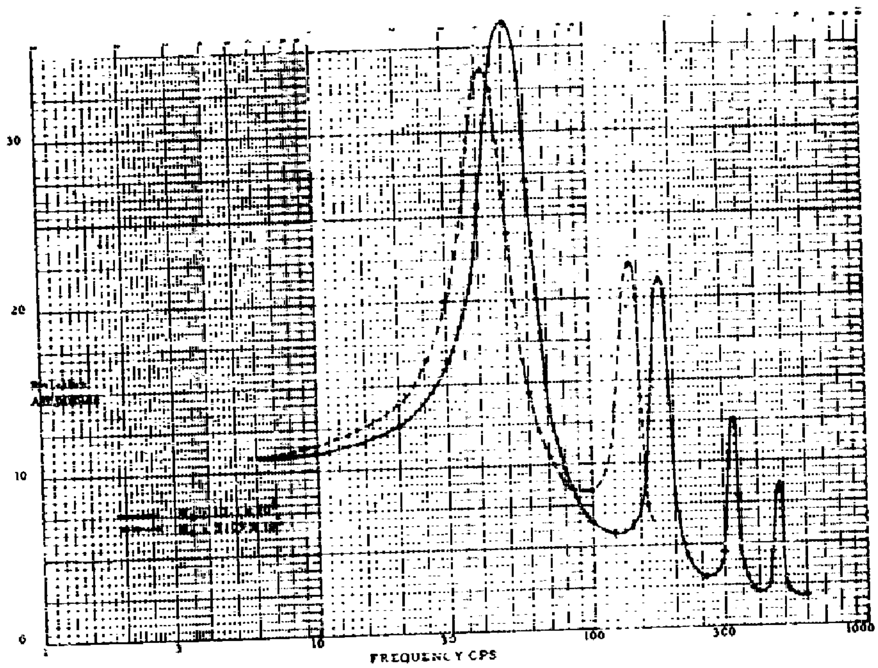


Figure 27 - Calculated Response of Transducer, Fittings, and Catheter as Spring Constant Per Unit Length of Catheter, K_d, is Varied.

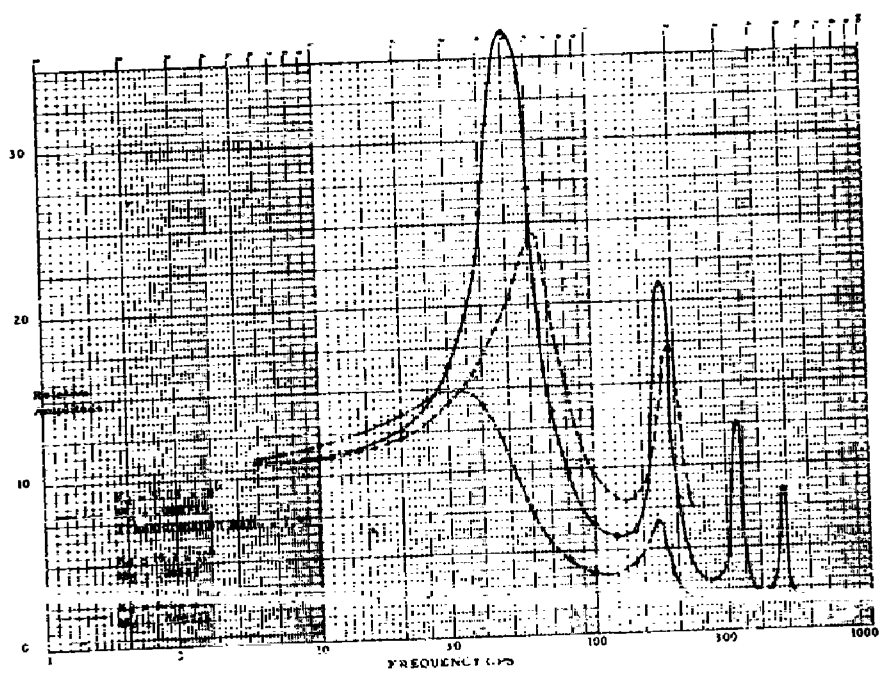


Figure 28 - Calculated Response of Transducer, Fittings, and Catheter as Spring Constant and Mass Per Unit Length, K_d and M_d, are Varied and as Spring Constant and Mass Per Unit Length, K_d and M_d, and Transformation Ratio are Varied.

resonant frequencies, (for K_d one half its normal value), represents the effect of making the catheter with a thinner wall or making it from a more pliable material. If one desires to use a catheter with a smaller cross sectional area, it would, in all likelihood, mean a thinner and more pliable wall, i.e., a lower value of K_d. It would also mean a smaller value of m_d, and a considerably lowered value of the transformation ratio arising from the difference in diameter of the needle bore and the catheter lumen. The bottom curve of Figure 28 shows the theoretical response which might be expected from such a change. This curve illustrates the case where K_d and m_d have been halved. Halving m_d, however, means that the diameter has been reduced by a factor of $1/\sqrt{2}$. Hence, the transformation ratio has been reduced by a factor of one-quarter, since it is equal to the fourth power of the ratio of the diameters. This tremendous change in the response curve was brought about by merely halving K_d and reducing the diameter by 30 percent. It is thought that this effect is the most likely possibility for explaining the discrepancies between the theoretical response curve and the measured response curve of Figure 16. The middle curve of Figure 28 is an illustration of the correct

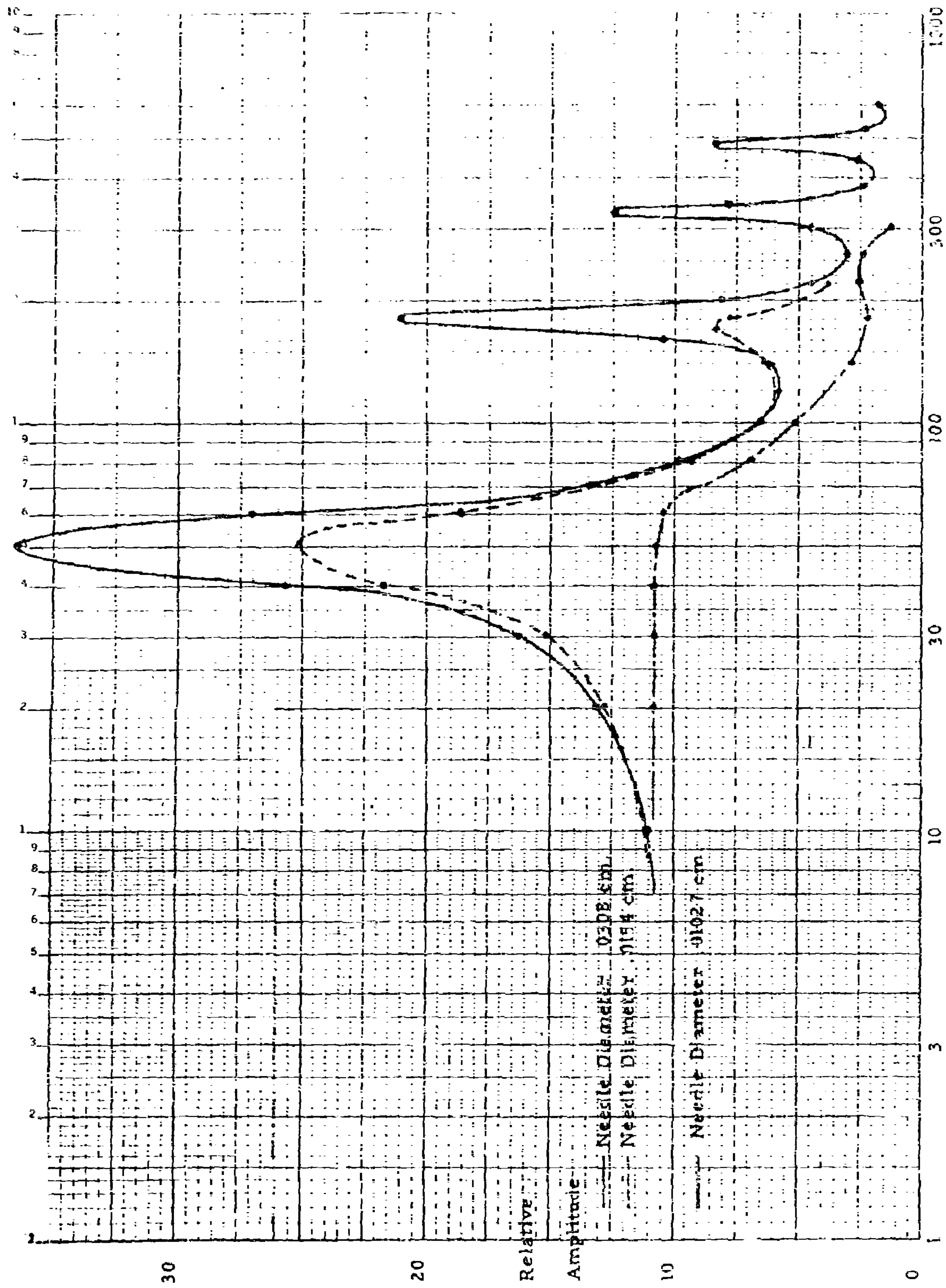


Figure 29 - Calculated Response of Transducer, Stopcock, Needle, and Catheter as Inside Diameter of Needle is Varied.

theoretical response curve if K_d had been measured erroneously as $10.1X10^6$ instead of $5.05X10^6$. (In this assumption, the velocity measurement was assumed to be correct, so m_d would be .000735 instead of .00147.)

The final effort in the variation of the system constants is illustrated in Figure 29. The curves illustrate what happens to the theoretical responses when the needle diameter is reduced. If the mass of the diaphragm and the mass of the liquid in the large diameter portions of the transducer are negligible, the mass m of the transducer is proportional to the square of the needle diameter. This condition is true in the Technitrol "Lilly" transducer used with a stopcock and No. 24 gauge needle since the needle diameter is much smaller than any of the larger diameter portions of the transducer. Thus, if the diameter is halved, the mass referred to the needle is reduced to one quarter of its former value. The above account of the transducer mass is for that mass referred to the needle diameter. If the mass referred to the catheter diameter is desired, the above value must be multiplied by the transformation ratio which is the fourth power of the ratio of the catheter

to the needle diameters. Thus, halving the needle diameter reduces the mass referred to the needle diameter to one-quarter of its former value, but it increases the mass referred to the catheter diameter to four times its former value, since

1/4 m new mass referred to needle	16 new transformation ratio	=	4m new mass referred to catheter	) for halved needle diameter. (48)

The behavior of the resistance for a reduction in needle diameter is not the same as the mass. In Equation (37)

$$R_d = -\frac{\frac{PA}{\ell}}{\frac{vol}{tA}} = -\frac{PA^2 t}{vol\,\ell} \quad . \tag{37}$$

This equation may be applied to the short needle by multiplying by 1, making the total resistance

$$R_d \ell = -\frac{PA^2 t}{vol} \equiv R_t \quad . \tag{49}$$

Now, if we assume streamline flow in the needle (which is justified from the data of Figures 22 and 23), the volume flow through the needle is from Poiseuille's Equation (31).

$$vol = \text{constant}\ \frac{d^4 P}{\ell} \tag{50}$$

If the diameter of the needle is halved, but the length is kept the same, the area is reduced to one-quarter of its former value, and the volume flow for a given pressure is reduced to one-sixteenth of its former value. Therefore, Equation (49) becomes:

$$R_t = -\frac{P\left(\frac{A}{4}\right)^2 t}{\frac{vol}{16}} = \frac{PA^2t}{vol} \tag{51}$$

for the needle with half of the original diameter. Thus, changing the needle diameter does not change the value of the resistance (referred to the changed needle diameter). However, it is to be noted, that the transducer resistance referred to the catheter diameter varies considerably. For example, if the needle diameter is halved, the transformation ratio is increased by a factor of 16 so the transducer resistance is increased by a factor of 16. In this analysis, as in the consideration of the mass, the resistance arising from the losses in the diaphragm or the liquid in the larger diameter portions of the transducer have been ignored. This procedure is justified since these losses are generally very small.

The value of K does not vary if one always considers it referred to the diameter of the catheter. This comes about because

$\frac{\Delta vol}{\Delta P}$ must be divided by the reciprocal of the square of the area of the needle bore in order to get $^1/K$ referred to the needle diameter. In order to get $^1/K$ referred to the catheter diameter, $^1/K$ referred to the needle diameter must be multiplied by the square of the ratio of the needle area to the catheter area. Thus, the needle area cancels out, and the value of K does not vary as the diameter of the needle is varied. The same argument applies to the mass and resistance associated with the diaphragm and the liquid in the larger diameter portions of the transducer if these quantities are not negligible. The result of the previous discussion leads to the equivalent circuit which follows

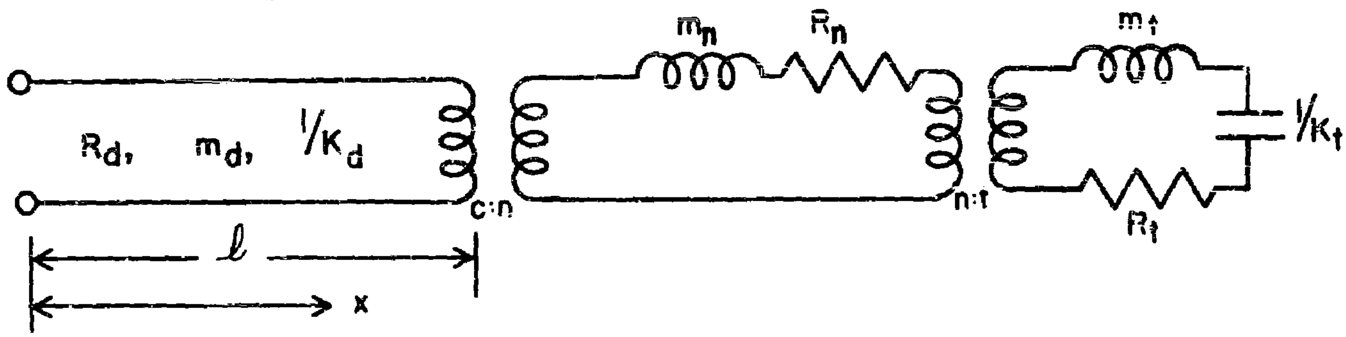


Figure 30 - Diagram of Electrical Analog of Transducer Catheter System, Including Transformations Arising From Changes in Internal Diameter,

where n subscript refers to the needle, and T subscript refers to the transducer. In this circuit, halving the diameter of the needle increases the catheter to needle and the transducer to needle trans-

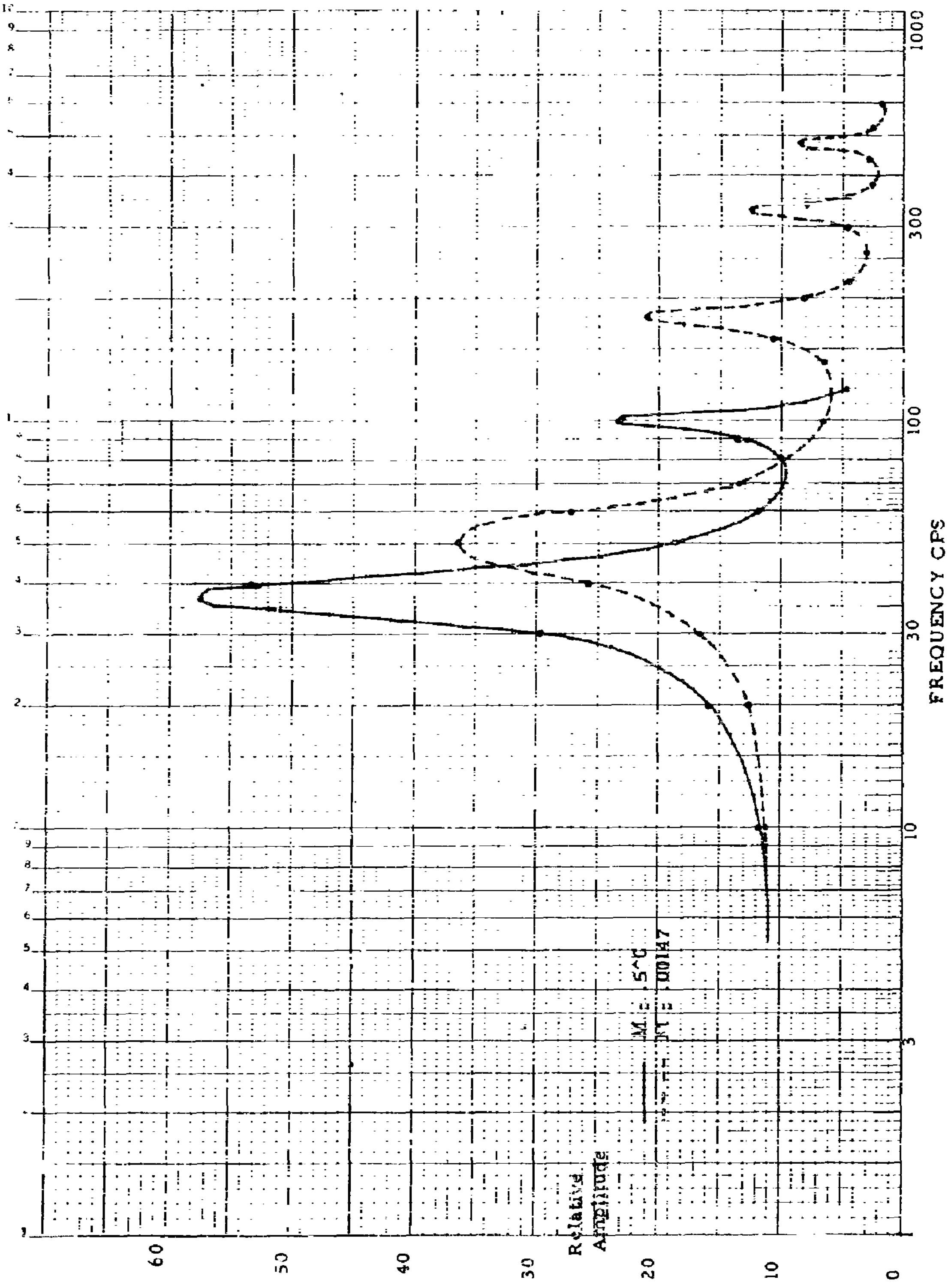


Figure 31 - Calculated Response of Transducer, Fittings, and Catheter as the Transducer Mass, M, is Increased to the Value Which Causes the Transducer to Resonate at 50 cps.

formation ratio by a factor of 16, and decreases the mass of the needle by a factor of 4. All of the other quantities remain the same.

If the length of the needle is quadrupled and the diameter is held constant, from Equation (49) it is seen that the resistance is increased by a factor of 4. The transformation ratio is the same, so the mass for this case is the same as for the case where the diameter was halved (Equation (48)). Thus, it is seen that reducing the cross sectional area of the needle is a much more effective way to increase the transducer resistance than increasing the needle length. It is to be noted that either of these methods of increasing the transducer resistance without appreciably lowering the frequency of the first resonant peak of the system are usable only if the resonant frequency of the transducer alone is considerably above this first system resonant peak. To illustrate this, a hypothetical transducer mass has been chosen such that the transducer itself resonates at 50 cps, and the response of the transducer-catheter system calculated. Figure 31 shows this response as well as the response of the normal system without

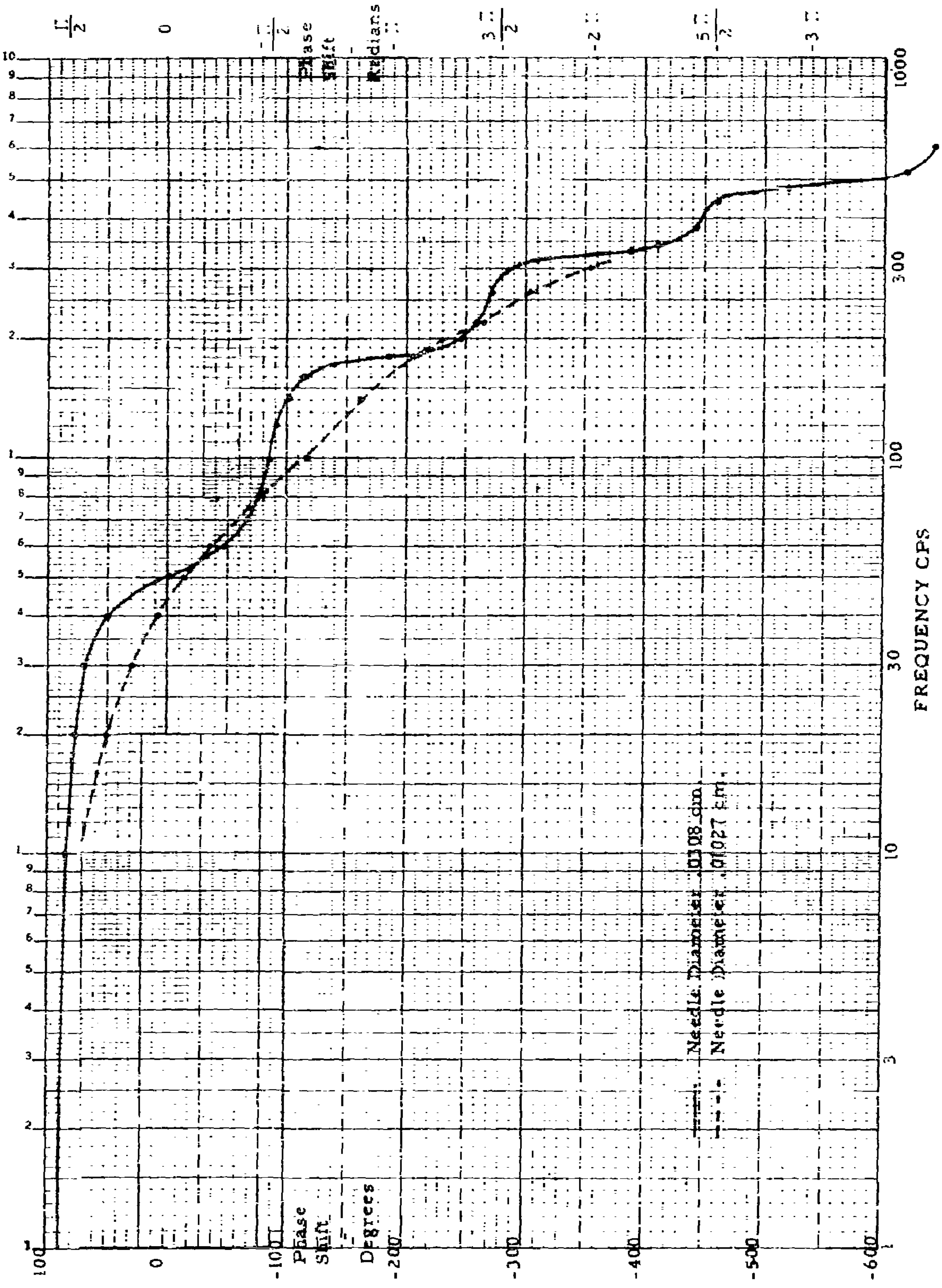


Figure 32 - Calculated Phase Response of Transducer, Stopcock, Needle, and Catheter as Inside Diameter of Needle is Varied.

alteration. These curves indicate that decreasing the needle diameter would increase the transducer mass further, and would lower the frequency of the first system resonance to an undesirable degree.

The effect of increasing damping by decreasing needle diameter is in agreement with the statements of Hansen (6). It should be pointed out, however, that the resonant frequency of the transducer and fittings (including needle) alone must be well above the first resonance of the transducer-fittings-catheter system.

As a matter of interest, the calculated phase angle versus frequency curves for two needle sizes (corresponding to the upper and lower amplitude-frequency curves of Figure 29) have been plotted on Figure 32. As would be expected, the underdamped case exhibits rather abrupt changes in the phase-frequency response while in the nearly critically damped case, the phase-frequency response is almost smooth.

E. TRANSIENT RESPONSE OF SYSTEM

The analysis of the distributed system has been carried out using the response to a pressure sine wave, i.e., the amplitude-frequency response evaluation. The use of this evaluation instead of the response of the distributed system to a pressure step function is dictated by the complexity of the analysis of the latter. Now that the amplitude-frequency response has been obtained, however, the response to the step function can, at least, be discussed qualitatively. Figure 8 shows the response to a pressure step function of the system consisting of the Technitrol "Lilly" transducer, fittings, and the 258 cm. No. 19 gauge catheter. The period of the predominant oscillation is about .025 seconds, while that of the secondary oscillation is about .00625 seconds. These periods correspond to the frequencies 40 and 160 cps, or almost exactly the frequencies of the first two resonances of the measured amplitude-frequency curve. One may also see from Figure 32 that the secondary oscillation amplitude is considerably smaller than the primary one, which is also in agreement with the measured amplitude-frequency curve. Presumably, the third and

fourth resonances discernible on the measured amplitude-frequency curve are too highly damped to be noticeable on the step function response.

F. APPLICATION OF THEORETICAL CONCEPTS

The practical value to the physiologists of such theoretical considerations of transducer systems as have been treated in this study is to be realized by the application of these concepts to mensuration techniques in use at present.

The effect of changing certain variables of the distributed transducer system upon the amplitude versus frequency response of the system has been pointed out from the theoretical calculations enumerated in Parts D and E of Phase II.

Since the purpose in transducer system design is to reproduce a pressure variable with minimal amplitude and phase distortion and with maximal frequency response, the desirability of developing a system which has minimal overshoot, i.e., which is critically damped, is apparent. Therefore, consideration of the effect of changing certain variables is appropriate.

The effects of changing the following variables has been calculated when:

(1) Spring constant K of transducer is varied (Figure 24)

(2) Resistance of transducer K is varied (Figure 25)

(3) Resistance per unit length of catheter is varied (Figure 26)

(4) Spring constant K_d per unit length of catheter, (Figure 27) K_d is varied

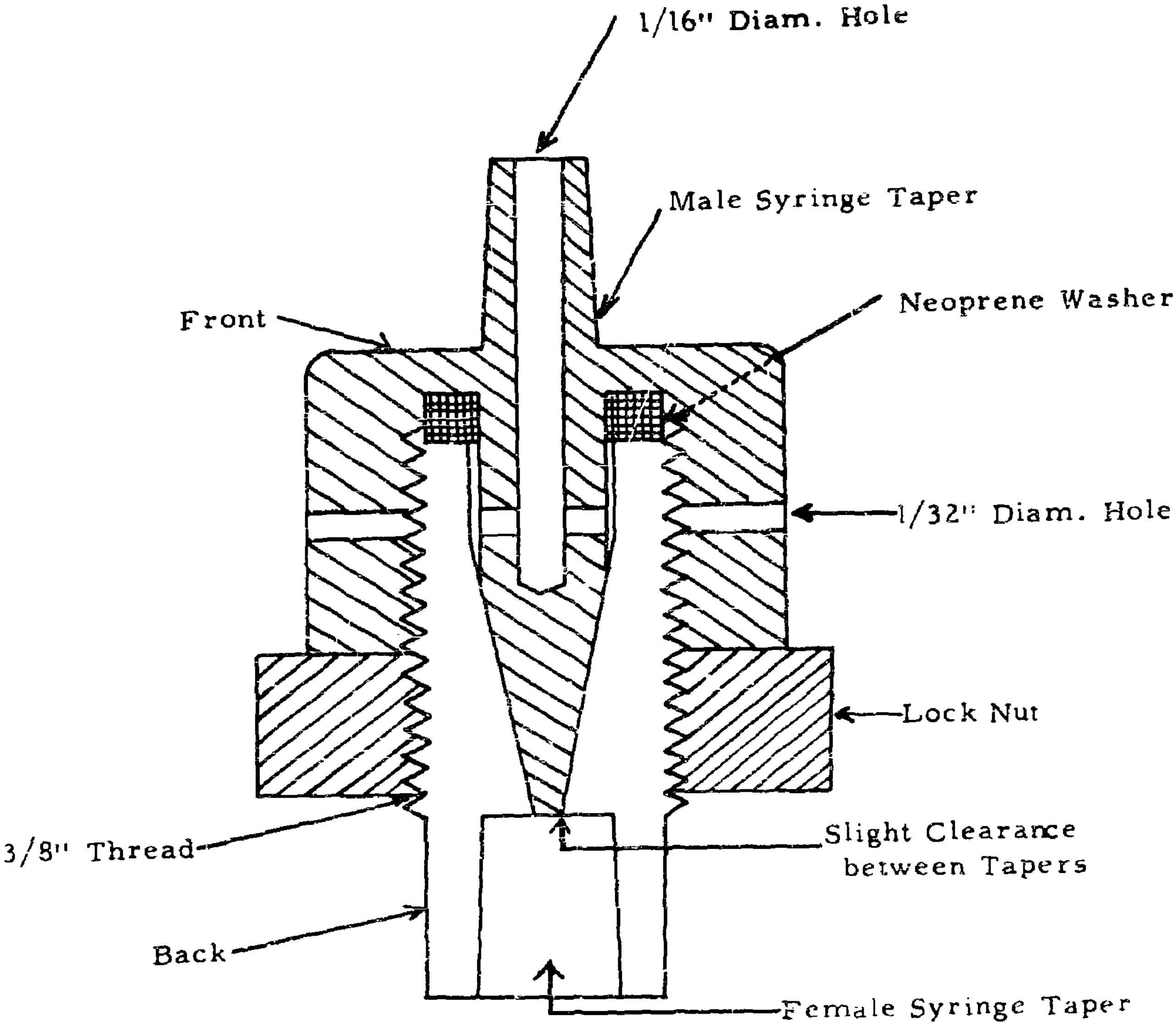
(5) Spring contant and mass per unit length of catheter K_d and m_d are varied (Figure 28)

(6) Spring constant, K_d, mass per unit length m_d and the transformation ratio are varied (Figure 28)

(7) Inside diameter of the needle is varied (Figure 29)

(8) The transducer mass m is increased to the value which causes the transducer to resonate at 50 cps, (Figure 30).

From the lower curves of Figure 29, it is seen that any short liquid transmission segment with a small but adjustable cross section may be used to vary the damping of a pressure measur-



Assembly has circular symmetry except for
1/32 inch damping hole
Scale - 4 times actual size
Material - Stainless steel except for neoprene washer

Figure 33 - Diagram of Suggested Device to Obtain Adjustable Damping.

ing system provided that the resonant frequency of the system does not become too low.

Such a damping segment was designed (Figure 33). This device has been incorporated into a transducer system with a large amplitude distortion at the natural frequency and has been found to be effective in providing adjustable damping in producing a frequency response curve which is essentially linear throughout. The results of these tests are shown in Figure 34. This unit has been found entirely practical in affecting variable damping of distributed and lumped transducer systems with a minimal decrease in the range of the frequency response.

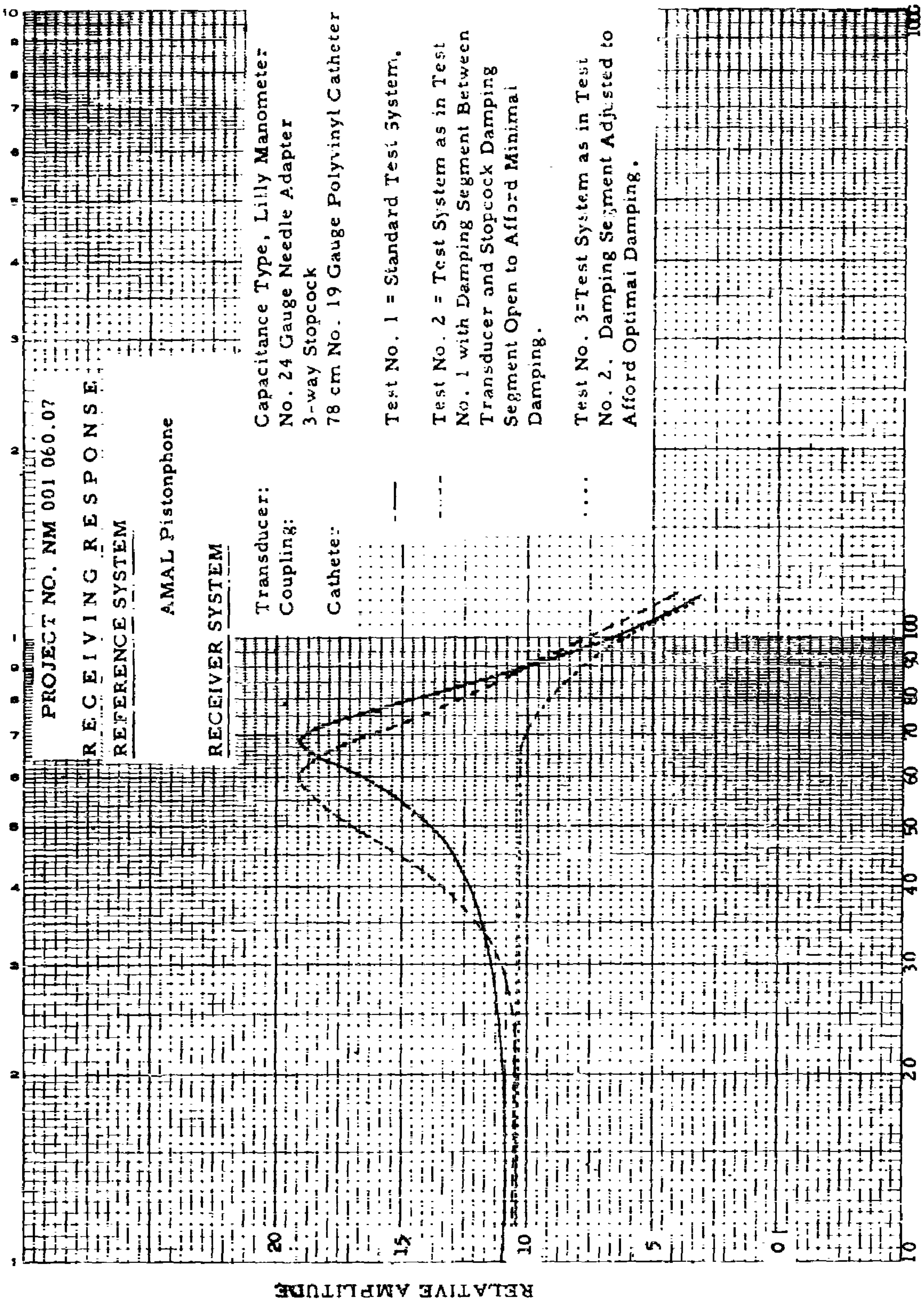


Figure 34 - Comparative Frequency Response Curves to Show Effectiveness of Damping Segment.

CONCLUSIONS

1. The step function and amplitude versus frequency methods of testing and evaluating lumped pressure measuring systems have been perfected to a high degree, and are in excellent agreement with each other as well as with the results of tests conducted with absolute calibration techniques.

2. The amplitude-frequency method of testing and evaluating distributed systems using the pistonphone is as valid as it is when used with lumped systems.

3. Because of the complexity of interpretation, the response of distributed systems to a pressure step function is only a qualitative, but very useful indication of the system response.

4. The theoretical treatment of the distributed system using constants measured at static pressure (and by other means) is not entirely a description of the behavior of a transducer-catheter system, but it is quite adequate for indicating trends when the physical constants and geometry of the system are varied.

5. A variable damping segment has been developed for use in increasing the useable portion of the amplitude versus frequency response of transducer systems.

RECOMMENDATIONS

1. In employing transducer systems, it is essential that they be tested for proper operation and response before and after measurements are taken.

2. The apparatus described herein and in Appendix B is recommended as adequate and easily constructed evaluation equipment.

3. Further transducer-catheter investigations should consider the concept of the analog of the electrical transmission line as described in this work.

4. The use of a damping segment such as the one previously described is recommended for producing variable damping in a transducer system.

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TABLE I

WARBURG'S TABLE NO. I FOR DETERMINATION OF R/m AND K/m OF CRITICALLY DAMPED AND OVERDAMPED SYSTEMS

Values of A for different values of α and τ.

τ \ α	0.2	0.4	0.6	0.8	1.0	1.2	1.4	1.6	1.8	2.0
0.5	0.1148									
1.0	0.4052	0.3528	0.3224	0.2908	0.2642	0.2416	0.2223	0.2052	0.1854	0.1778
1.5	0.7751	0.6579	0.5686	0.4982	0.4422	0.3968	0.3598	0.3283	0.3020	0.2795
2.0	1.1279	0.9272	0.7831	0.6759	0.5940	0.5296	0.4781	0.4352	0.3998	0.3697
2.5	1.3378	1.1221	0.9410	0.8236	0.7127	0.6367	0.5762	0.5259	0.4841	0.4487
3.0	1.5151	1.2281	1.0382	0.9028	0.8009	0.7208	0.6563	0.6021	0.5567	0.5178
3.5	1.5049	1.2532	1.0847	0.9607	0.8641	0.7859	0.7214	0.6661	0.6191	0.5782
4.0	1.3844	1.2189	1.0946	0.9933	0.9084	0.8361	0.7743	0.7199	0.6727	0.6311
4.5	1.2008	1.1518	1.0826	1.0091	0.9389	0.8746	0.8171	0.7651	0.7188	0.6774
5.0	1.0055	1.0760	1.0608	1.0147	0.9596	0.9041	0.8518	0.8028	0.7583	0.7178
5.5	0.8446	1.0100	1.0377	1.0147	0.9734	0.9266	0.8799	0.8345	0.7924	0.7532
6.0	0.7473	0.9637	1.0180	1.0122	0.9826	0.9439	0.9027	0.8612	0.8216	0.7842
6.5	0.7237	0.9400	1.0039	1.0091	0.9887	0.9571	0.9211	0.8835	0.8467	0.8113
7.0	0.7657	0.9362	0.9955	1.0061	0.9927	0.9672	0.9361	0.9023	0.8683	0.8349
7.5	0.8521	0.9465	0.9917	1.0038	0.9953	0.9749	0.9482	0.9180	0.8868	0.8556
8.0	0.9556	0.9643	0.9911	1.0021	0.9970	0.9808	0.9580	0.9312	0.9027	0.8737
8.5	1.0503	0.9833	0.9924	1.0010	0.9981	0.9853	0.9660	0.9423	0.9164	0.8896
10.0						0.9934	0.9819	0.9659	0.9470	0.9261

TABLE II

WARBURG'S TABLE NO. II FOR DETERMINATION OF R/m AND K/m OF CRITICALLY DAMPED AND OVERDAMPED SYSTEMS

α	$\tau_{A_{0.4}}$	$\tau_{A_{0.9}}$	$\frac{\tau_{A_{0.4}}}{\tau_{A_{0.9}}}$	α	$\tau_{A_{0.4}}$	$\tau_{A_{0.9}}$	$\frac{\tau_{A_{0.4}}}{\tau_{A_{0.9}}}$
0.20	0.99	1.68	0.589	1.15	1.48	4.67	0.317
0.25	1.01	1.74	0.580	1.20	1.51	4.93	0.306
0.30	1.03	1.80	0.572	1.25	1.55	5.18	0.299
0.35	1.05	1.87	0.561	1.30	1.59	5.44	0.292
0.40	1.07	1.95	0.549	1.35	1.63	5.69	0.286
0.45	1.09	2.04	0.534	1.40	1.67	5.94	0.281
0.50	1.11	2.14	0.519	1.45	1.71	6.19	0.276
0.55	1.14	2.25	0.507	1.50	1.75	6.44	0.272
0.60	1.16	2.37	0.490	1.55	1.79	6.69	0.268
0.65	1.18	2.50	0.472	1.60	1.83	6.94	0.264
0.70	1.21	2.64	0.458	1.65	1.87	7.19	0.260
0.75	1.23	2.80	0.439	1.70	1.91	7.43	0.257
0.80	1.26	2.98	0.423	1.75	1.96	7.67	0.256
0.85	1.29	3.19	0.404	1.80	2.00	7.92	0.253
0.90	1.32	3.42	0.386	1.85	2.04	8.17	0.250
0.95	1.35	3.66	0.369	1.90	2.09	8.42	0.248
1.00	1.38	3.91	0.353	1.95	2.14	8.67	0.247
1.05	1.41	4.16	0.339	2.00	2.19	8.93	0.245
1.10	1.44	4.42	0.326				

APPENDICES

APPENDIX A

DEFINITION OF SYMBOLS

With the exception of terms used once and defined at the place used, all of the symbols used in the text are defined in the following table:

1. A - area
2. C - electrical capacitance
3. C_d - electrical capacitance per unit length (of a transmission line)
4. d - diameter
5. E - sinusoidal driving voltage amplitude
6. F - force
7. f_n - natural frequency
8. f_{nd} - damped natural frequency
9. G_d - electrical conductance per unit length (of a transmission line)
10. i - electrical current
11. $\underline{i}$ - Laplace transform of electrical current
12. K - spring constant (from Hook's law $F = KX$)
13. K_d - spring constant per unit length (of a mechanical transmission line)

14. L - electrical inductance

15. L_d - electrical inductance per unit length

16. l - length

17. m - mass

18. m_d - mass per unit length (of a mechanical transmission line)

19. p - Laplace transform of the first time derivative $\left(\mathcal{L}\left\{\frac{d}{dt}\right\}\right)$

20. P - pressure

21. q - electrical charge

22. R - mechanical resistance (coefficient of viscous friction)

23. R_d - mechanical resistance per unit length (of a mechanical transmission line.)

24. R_e - electrical resistance

25. R_{ed} - electrical resistance per unit length (of a transmission line)

26. r - radius

27. S - amplitude of a step function

28. t - time

29. V - velocity

30. v -electrical voltage

31. $\underline{v}$ - Laplace transform of electrical voltage

32. vol - volume

33. x - mechanical position

34. Y - sinusoidal driving force amplitude

35. Z_r - terminating impedance of the receiving end of a transmission line

36. Z_c - characteristic impedance of a transmission line

37. γ - propagation constant (either electrical or mechanical)

38. β - imaginary part of γ

39. Δ - increment (a small change in the quantity with which it is used)

40. ω - angular frequency in radians of a driving quantity (voltage or force)

41. ϕ - phase angle

42. ω_n - natural angular frequency in radians

43. ω_{nd} - damped natural angular frequency in radians

44. λ - wavelength

45. I - sinusoidal response current amplitude

NOTE: The centimeter-gram second system of units is used except where otherwise noted.

APPENDIX B

TRANSDUCER SYSTEMS AMPLITUDE VERSUS FREQUENCY RESPONSE CURVES

TRANSDUCER		COUPLING					NEEDLE	TEST SYSTEM	Page in Appendix B
Type	Make	Stopcock	Fitting	Type	Catheter Size	Length			
Variable Capacitance	Lilly	None	None	None	None	None	None	USRL Free Field Reciprocal Calibration System	1
Variable Capacitance	Lilly	3-way Stopcock Integrated into Barrel Adapter	TB Syringe Barrel Adapter	Polyvinyl	No. 21 Ga. 0.3 mm. I.D.	12 cm.	None	Step Function	2
"	"	" "	" "	"	" "	" "	"	AMAL Pistonphone	2
Variable Capacitance	Lilly	3-way Stopcock	No. 24 Ga. Needle Adapter	Polyvinyl Barium Impregnated	No. 19 Ga. 0.45 mm. I.D.	77 cm.	None	Step Function	3
"	"	" "	" "	" "	" "	" "	"	AMAL Pistonphone	3
Variable Capacitance	Lilly	3-way Stopcock	No. 24 Ga. Needle Adapter	Polyvinyl Barium Impregnated	" "	90 cm.	"	USRL Free Field Reciprocal Calibration	4
"	"	" "	" "	" " Note: Same as above, but with 80 cm. of catheter enclosed in 2 mm. I.D., 4 mm. O.D. steel tubing.	" "	" "	"	" " "	4
Variable Capacitance	Lilly	3-way Stopcock	No. 24 Ga. Needle Adapter	Polyvinyl Barium Impregnated	No. 19 Ga. 0.45 mm. I.D.	188.5 cm.	None	Step Function (Calculated - single degree of freedom)	5

APPENDIX B - CONTINUED

TRANSDUCER SYSTEMS AMPLITUDE VERSUS FREQUENCY RESPONSE CURVES

TRANSDUCER		COUPLING					NEEDLE	TEST SYSTEM	Page in Appendix B
Type	Make	Stopcock	Fitting	Type	Catheter Size	Length			
Variable Capacitance	Lilly	3-way Stopcock	No. 24 Ga. Needle Adapter	Polyvinyl Barium Impregnated	No. 19 Ga. 0.45 mm. I.D.	275 cm.	None	USRL Low Frequency Tank	6
Variable Capacitance	Lilly	3-way Stopcock	No. 24 Ga. Needle Adapter	Polyvinyl Barium Impregnated	No. 19 Ga. 0.45 mm. I.D.	258 cm.	None	AMAL Pistonphone	7
Variable Capacitance	Lilly	3-way Stopcock	No. 28 Ga. Needle Adapter	Polyvinyl Barium Impregnated	No. 21 Ga 0.3 mm. I.D.	100 cm.	None	AMAL Pistonphone	[illegible]
"	"	" "	" "	"	" "	" "	"	USRL Low Frequency Tank	8
"	"	" "	" "	"	" "	" "	"	Step Function	8
Variable Capacitance	Lilly	3-way Stopcock	None	Cournand Cardiac Catheter	No. 4F	100 cm.	None	AMAL Pistonphone	9
Variable Capacitance	Lilly	3-way Stopcock	None	Cournand Cardiac Catheter	No. 7F	100 cm.	None	AMAL Pistonphone	10
Variable Capacitance	Sanborn	None	No. 24 Ga. Needle Adapter	Polyvinyl Zarium Impregnated	No. 19 Ga. 0.45 mm. I.D.	100 cm.	None	AMAL Pistonphone	11
"	"	"	" "	"	" "	" "	"	USRL Low Frequency Tank	11
"	"	"	" "	"	" "	" "	"	Step Function Response	11
Variable Capacitance	Sanborn	None	No. 24 Ga. Needle Adapter and 47 cm. ½" lead tubing	Polyvinyl Barium Impregnated	No. 19 Ga. 0.45 mm. I.D.	85 cm. (10 cm. in tank)	None	USRL Low Frequency Tank	12

APPENDIX B - CONTINUED

TRANSDUCER SYSTEMS AMPLITUDE VERSUS FREQUENCY RESPONSE CURVES

TRANSDUCER		COUPLING					NEEDLE	TEST SYSTEM	Page in Append. B
Type	Make	Stopcock	Fitting	Type	Catheter Size	Length			
Variable Resistance	Statham	3-way Stopcock	47 cm. ¼" O.D. Load Tubing	None	None	None	None	USRL Low Frequency Task	13
"	"	" "	" "	"	"	"	20 Ga. 100 mm.	" " " "	13
"	"	" "	" "	"	"	"	Same Needle cut to 18 mm.	" " " "	3
Variable Resistance	Statham	3-way Stopcock	3" USRL .098" I.D. Tubing	None	None	None	No. 20 Ga. 100 mm. in length	USRL Low Frequency Task	14
"	"	" "	" "	"	"	"	Above Needle cut to 37 mm.	" " " "	4
"	"	" "	" "	"	"	"	Same Needle cut to 18 mm.	" " " "	14
Variable Resistance	Statham	3-way Stopcock	47 cm. ¼" O.D. Load Tubing	None	None	None	20 Ga. 13 mm.	USRL Low Frequency Task	15
"	"	" "	" "	"	"	"	" "	Step Function	15
"	"	" "	" "	"	"	"	" "	ANAL Pistonphone	15
Variable Resistance	Statham	3-way Stopcock	Double Male Taper and 47 cm. ¼" O.D. Load Tubing	None	None	None	20 Ga. 100 mm.	ANAL Pistonphone	16
"	"	" "	" "	"	"	"	" "	USRL Low Frequency Task	16
"	"	" "	" "	"	"	"	" "	Step Function	16

APPENDIX B - CONTINUED

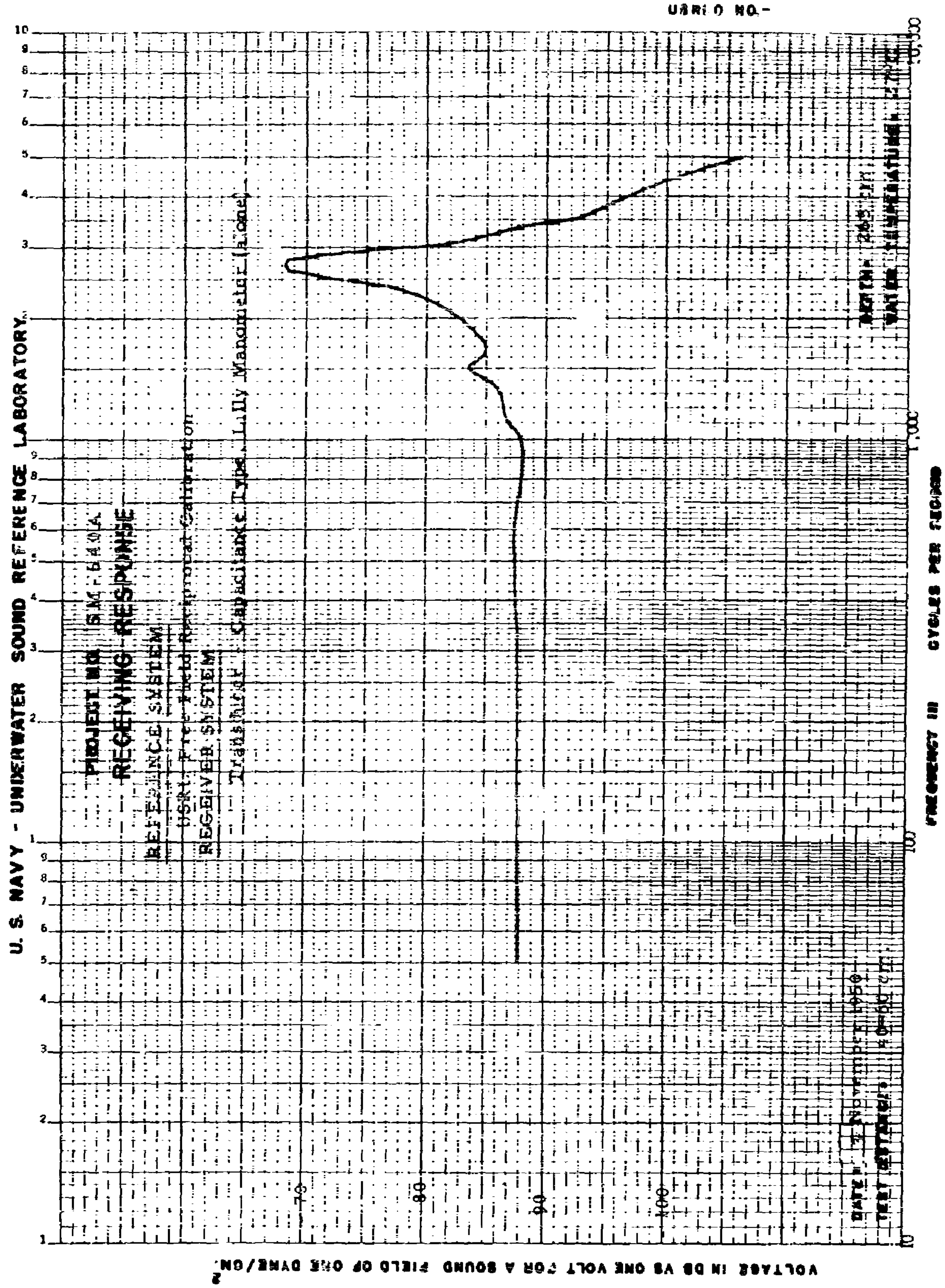
TRANSDUCER SYSTEMS AMPLITUDE VERSUS FREQUENCY RESPONSE CURVES

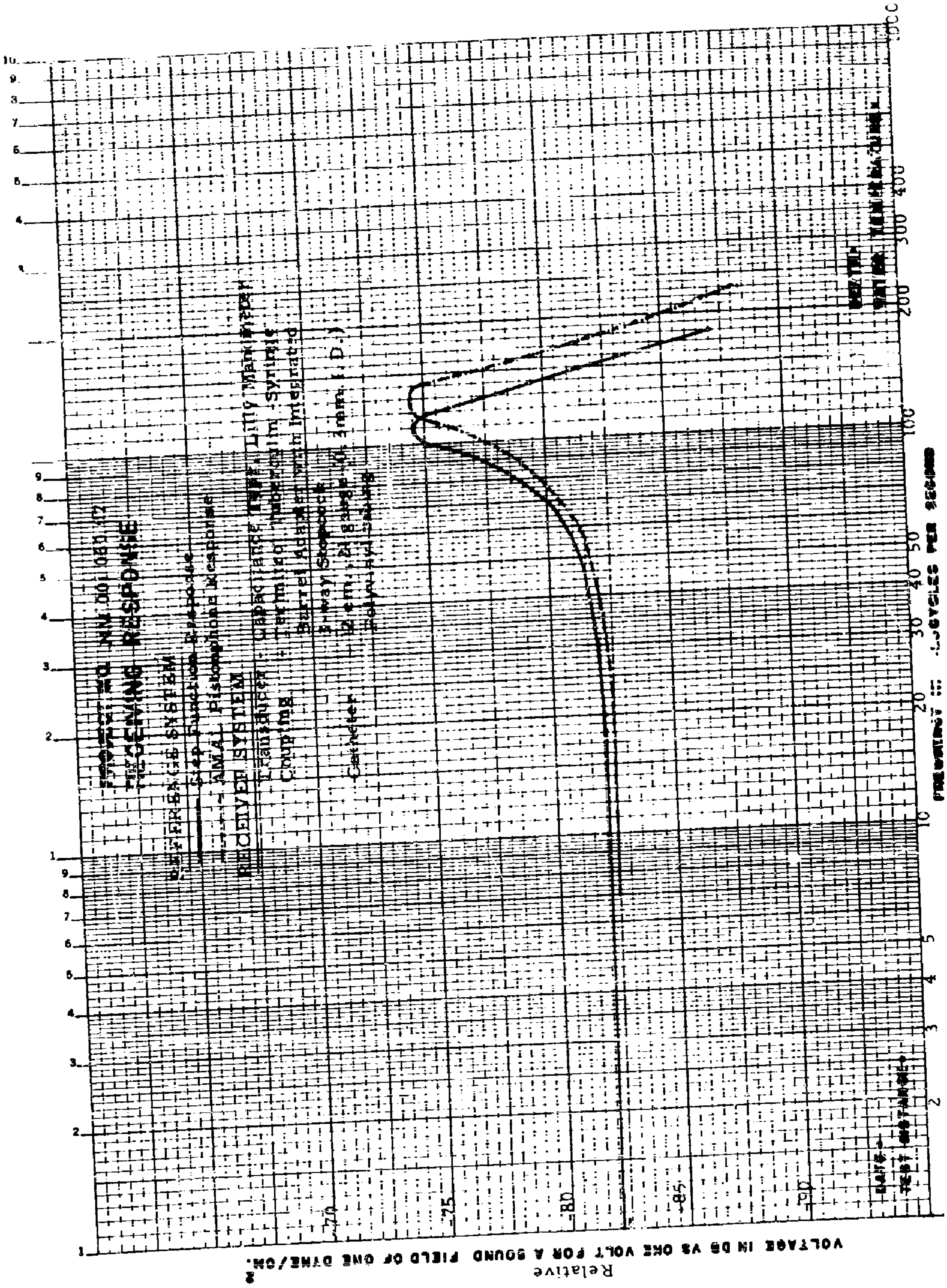
TRANSDUCER		COUPLING					NEEDLE	TEST SYSTEM	Page in Appendix B
Type	Make	Stopcock	Fitting	Type	Catheter Size	Length			
Variable Resistance	Statham	3-way Stopcock	Double Male Taper	Cournand Cardiac Catheter	No. 4F	100 cm. 6 cm. in test tank	None	USRL Low Frequency Tank	[illegible]
"	"	" "	" "	"	" "	Same as above, but with 90 cm. in tank	"	" " " "	[illegible]
Variable Resistance	Statham	3-way Stopcock	Double Male Taper	Cournand Cardiac Catheter	No. 7F Catheter No. 1	100 cm.	None	USRL Low Frequency Tank	[illegible]
"	"	" "	" "	"	No. 7F Catheter No. 2	"	"	" " " "	[illegible]
Variable Reluctance 3 mm. Replaceable Rubber	Gaeer	None	None	None	-	-	None	USRL Low Frequency Tank	[illegible]
Variable Reluctance 3 mm. Replaceable Rubber	Gaeer	None	None	None	-	-	None	AMAL Pistonphone	[illegible]

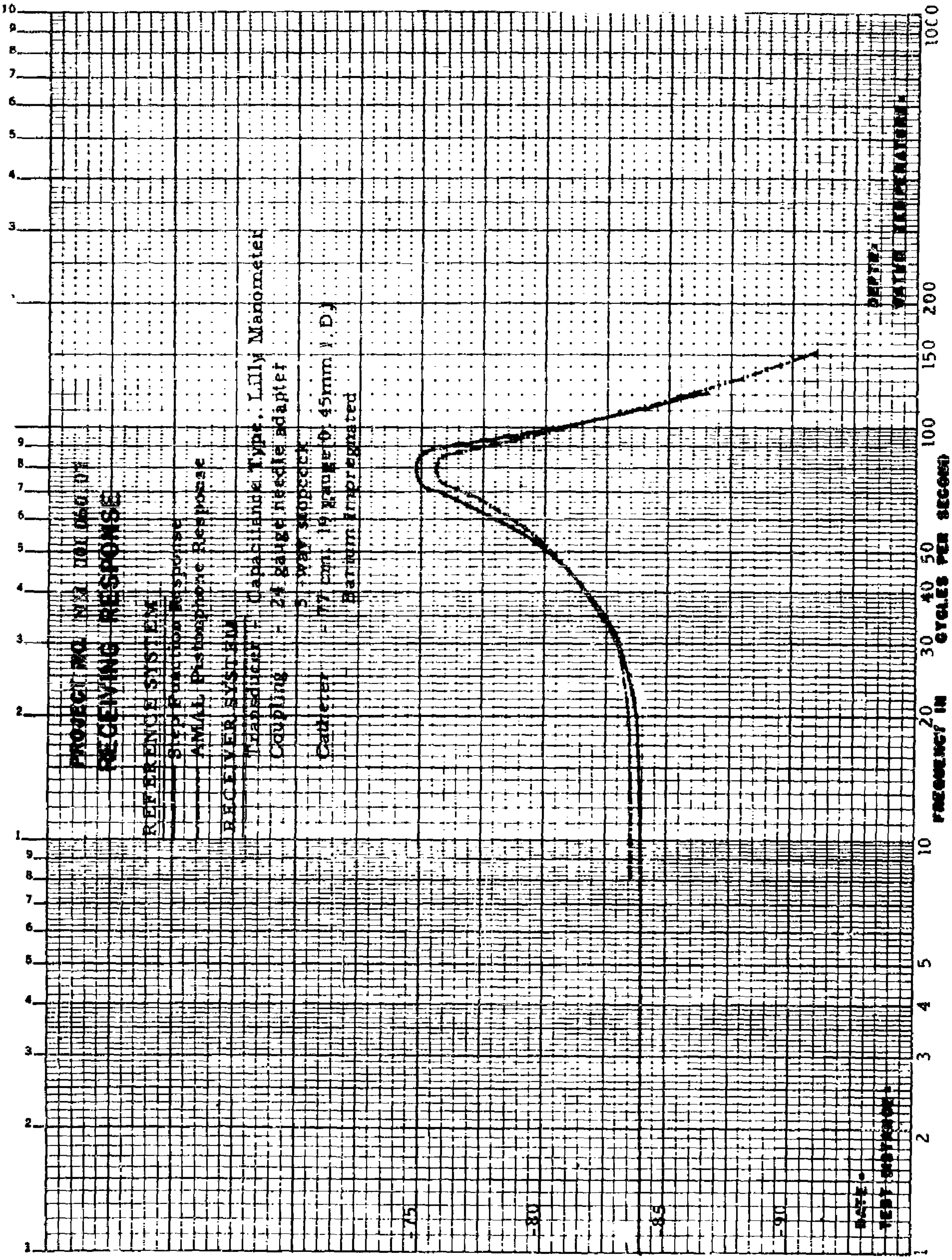
APPENDIX B - CONTINUED

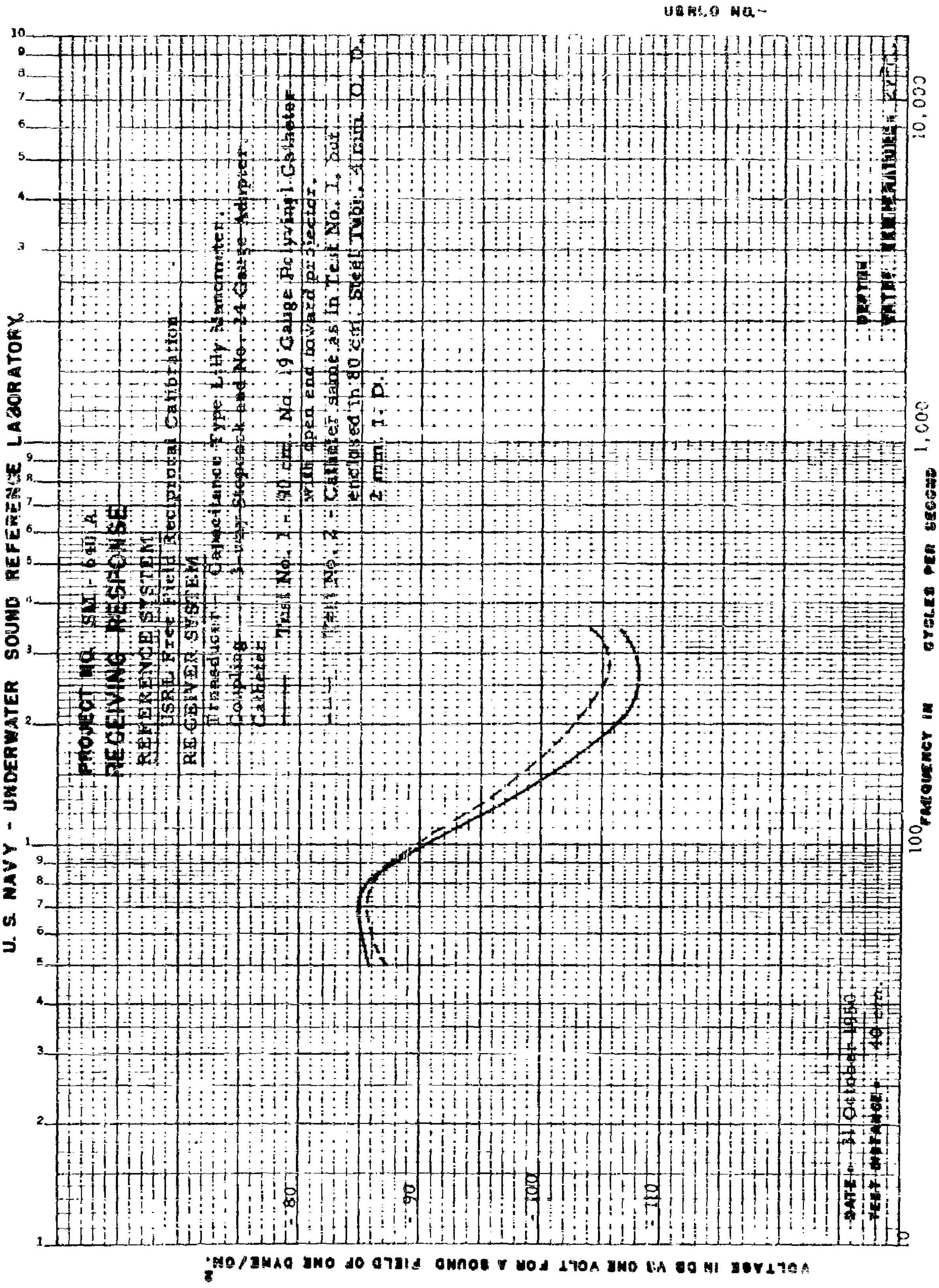
TRANSDUCER SYSTEMS AMPLITUDE VERSUS FREQUENCY RESPONSE CURVES

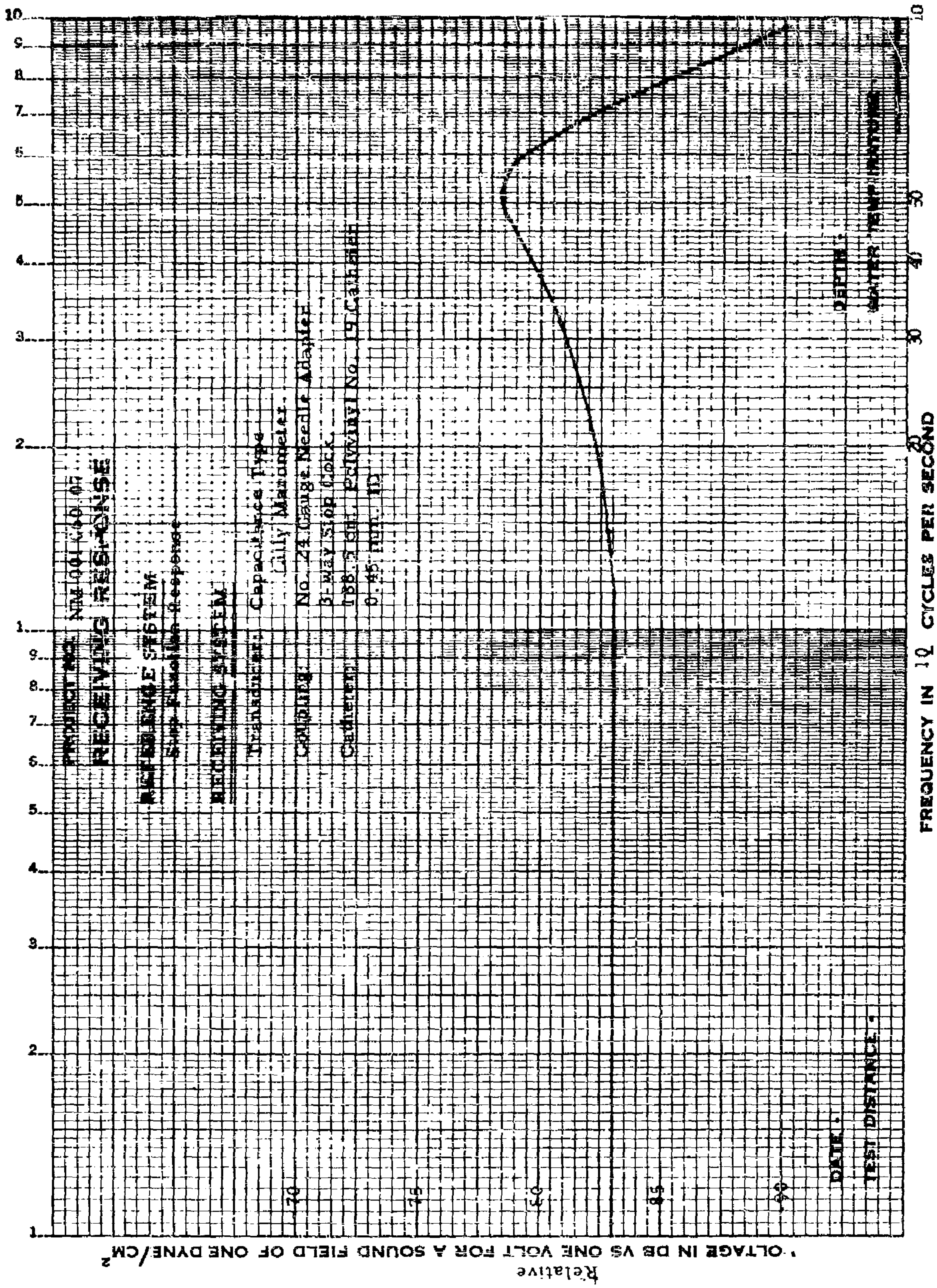
TRANSDUCER		COUPLING					NEEDLE	TEST SYSTEM	Page in Appendix B
Type	Make	Stopcock	Fitting	Type	Catheter Size	Length			
Variable Reluctance 3 mm. Replaceable Rubber	Gaeer	None	None	None	-	-	None	USRL Free Field Calibration	20
"	"	"	"	"	-	-	"	Step Function (Calculated)	20
Variable Reluctance 5 mm. Phos. Bronze Diaphragm	Gaeer	None	None	None	-	-	None	USRL Low Frequency Tank	21
Variable Reluctance 8 mm. Phos. Bronze Diaphragm	Gaeer	None	None	None	-	-	None	USRL Free Field Calibration	22

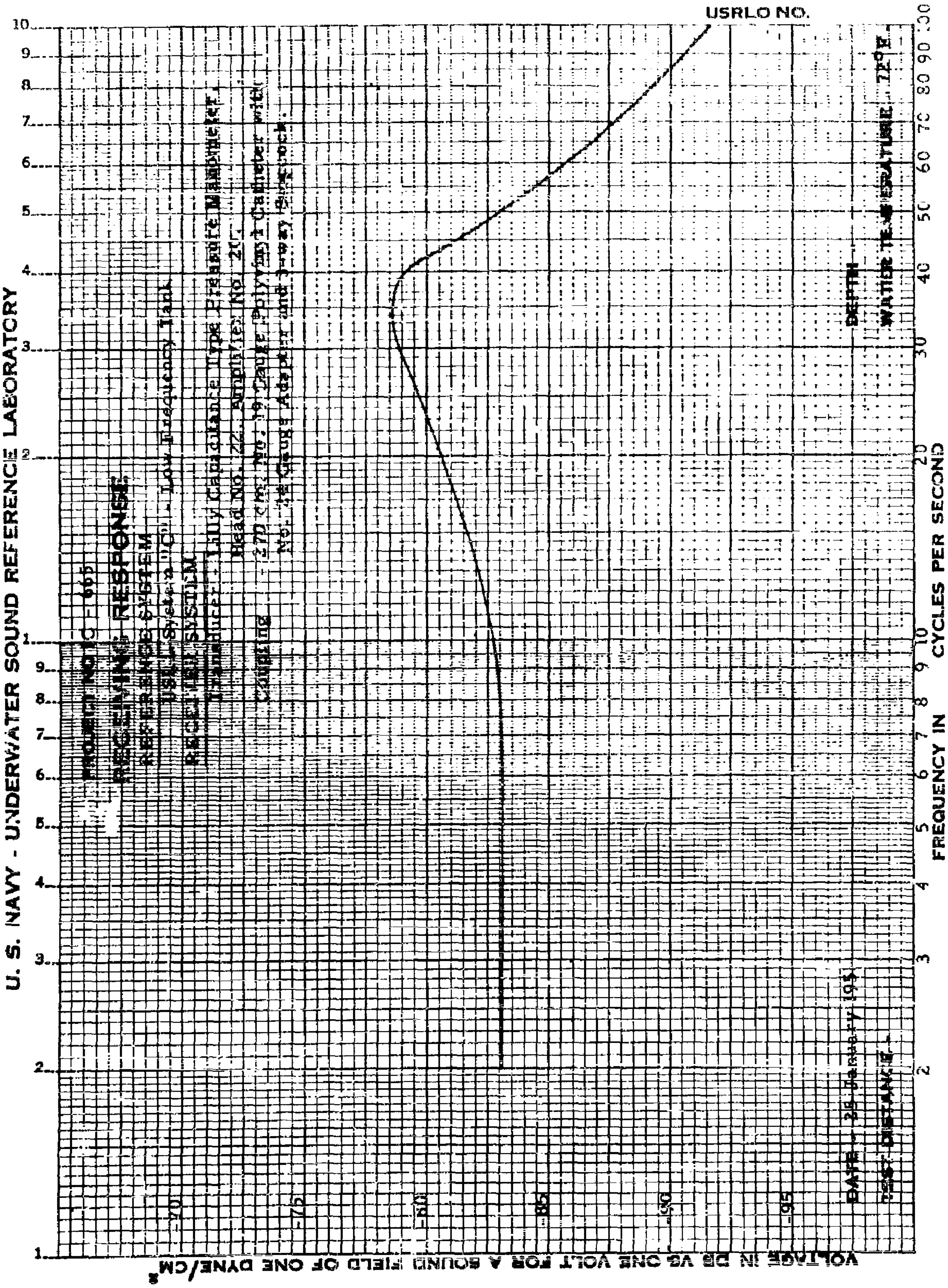

U. S. NAVY - UNDERWATER SOUND REFERENCE LABORATORY
RECEIVING RESPONSE
REFERENCE SYSTEM
RECEIVER SYSTEM
FREQUENCY IN CYCLES PER SECOND
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM²


RECEIVING RESPONSE
REFERENCE SYSTEM
RECEIVER SYSTEM
Transducer
Coupling
Catheter
FREQUENCY IN CYCLES PER SECOND
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM.²
Relative
-70
-75
-80
-85
-90


RECEIVING RESPONSE
REFERENCE SYSTEM
RECEIVER SYSTEM
Transducer - Capacitance Type, Lilly Manometer
Coupling - 24 gauge needle adapter
3 way stopcock
Catheter - 77 cm. 19 gauge 0.45mm I.D.
Barium Impregnated
-75
-80
-85
-90
FREQUENCY IN CYCLES PER SECOND
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM.2
Relative


USRL NO.
U. S. NAVY - UNDERWATER SOUND REFERENCE LABORATORY
PROJECT NO. SM-6(1) A
RECEIVING RESPONSE
REFERENCE SYSTEM
USRL Free Field Reciprocal Calibration
RECEIVER SYSTEM
Transducer — Capacitance Type L-Hy Manometer.
Coupling — 3-way Stopcock and No. 24 Gauge Adapter.
Catheter
Test No. 1 - 90 cm. No. 19 Gauge Polyvinyl Catheter with open end toward projector.
Test No. 2 - Catheter same as in Test No. 1, but enclosed in 80 cm. Steel Tube, 4 mm. O. D. 2 mm. I. D.
DATE - 31 October 1950
TEST DISTANCE - 40 cm.
DEPTH
-80
-90
-100
-110
100
1,000
10,000
FREQUENCY IN CYCLES PER SECOND
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM.²


RECEIVING RESPONSE
REFERENCE SYSTEM
RECEIVING SYSTEM
Transducer: Capacitance Type
Lilly Manometer
Coupling: No. 24 Gauge Needle Adapter
3-way Stop Cock
0.45 mm. ID
FREQUENCY IN 10 CYCLES PER SECOND
DEPTH:
DATE:
TEST DISTANCE:
Relative
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM²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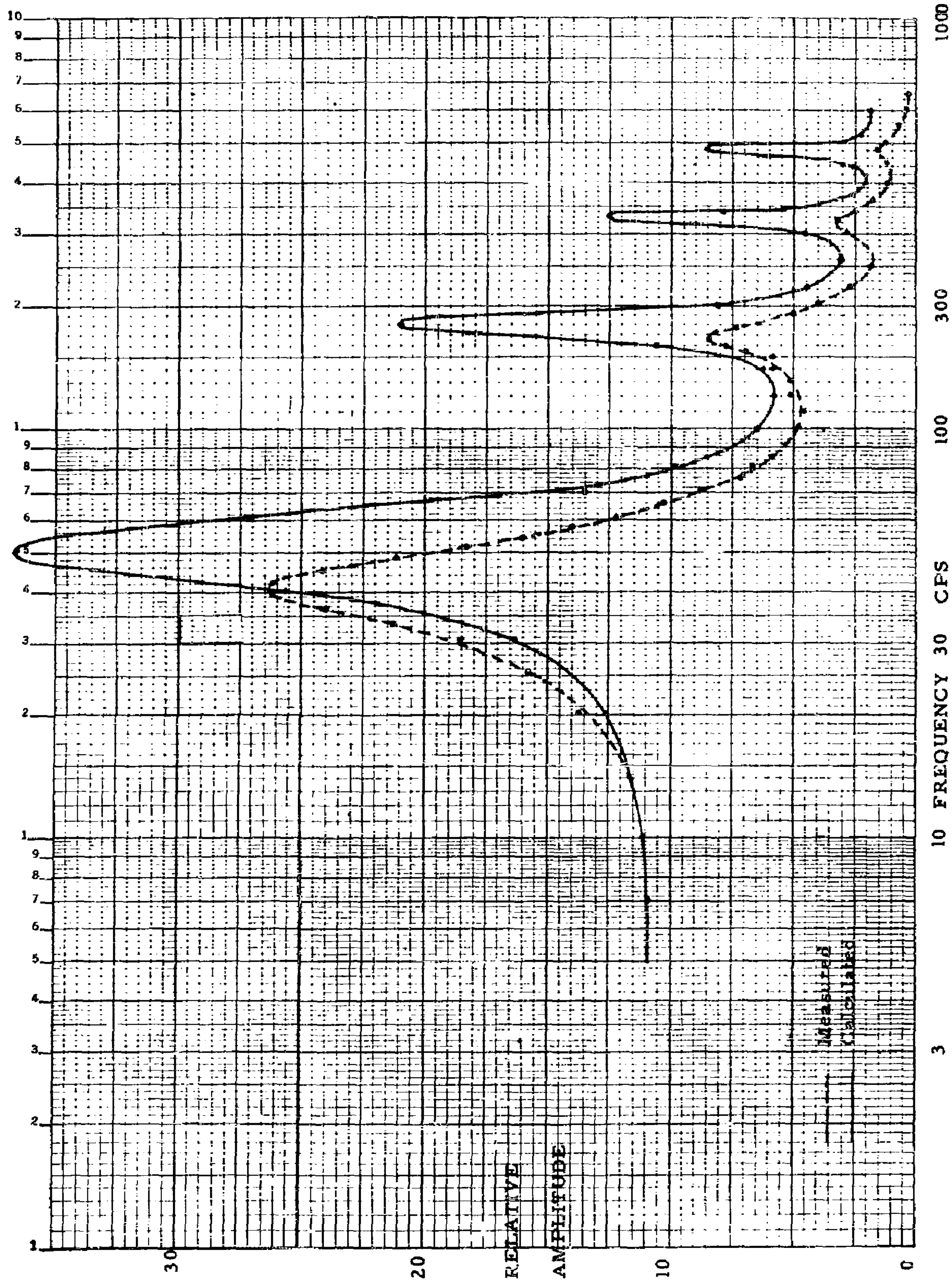
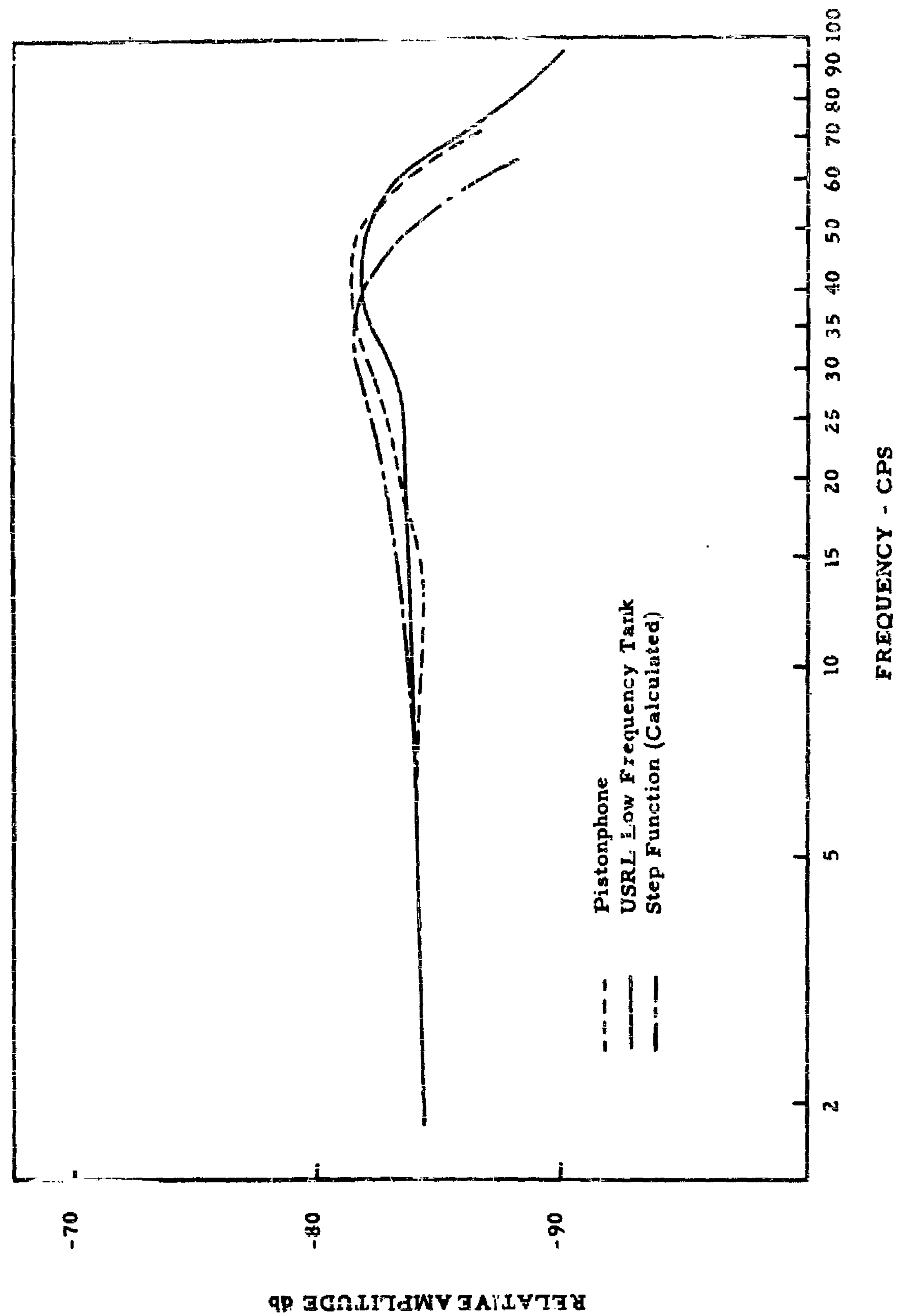
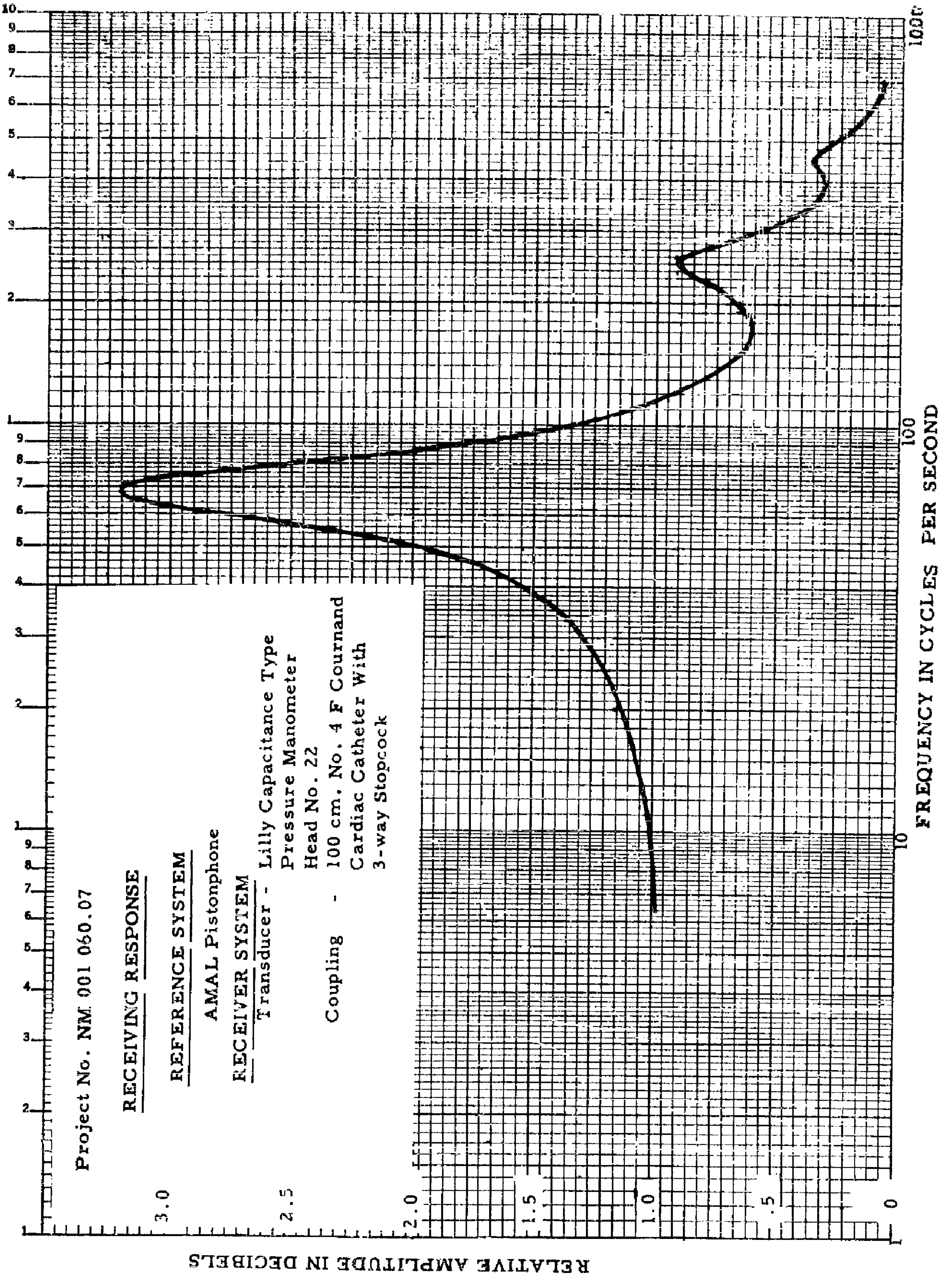
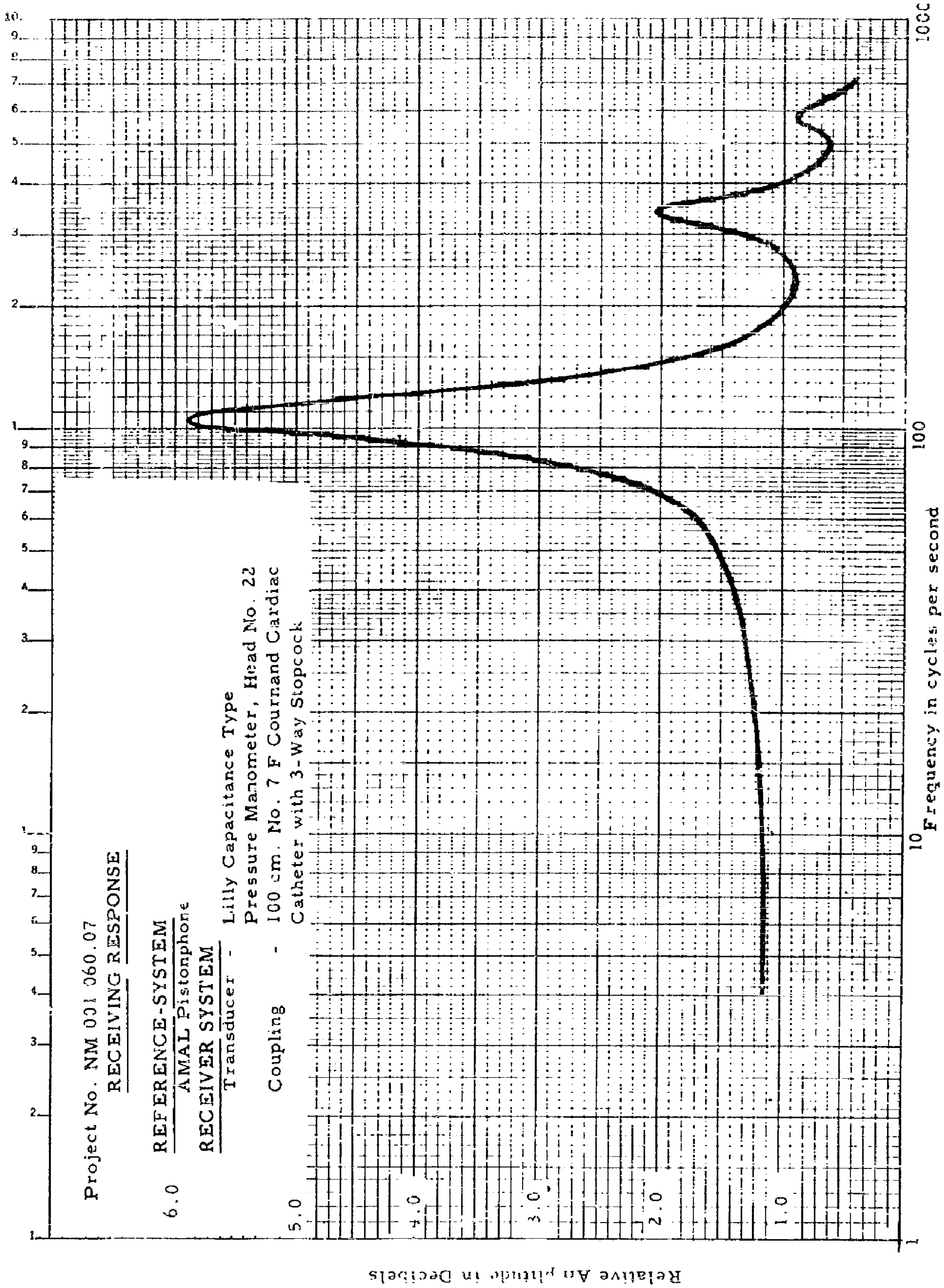
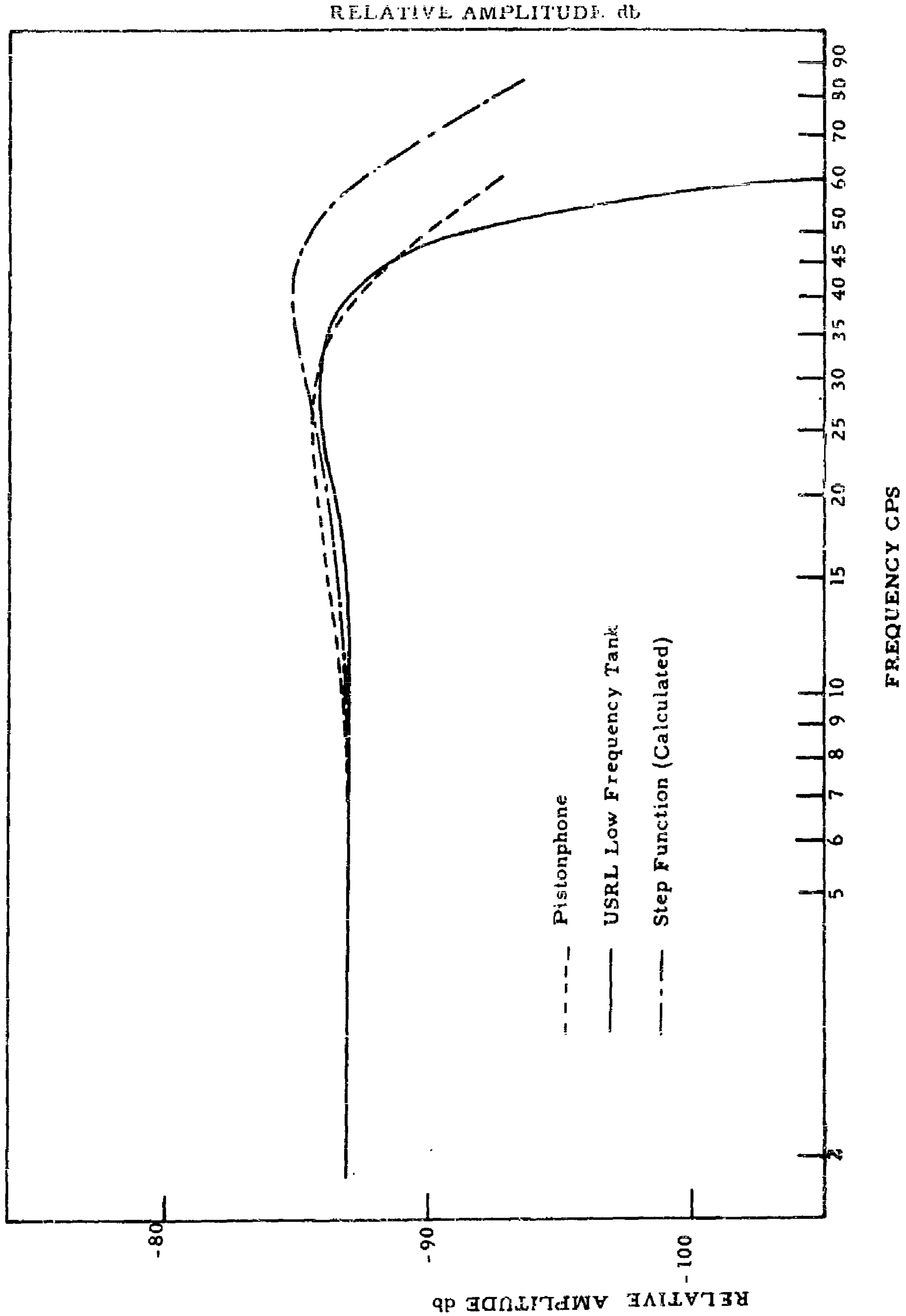
U. S. NAVY - UNDERWATER SOUND REFERENCE LABORATORY
PROJECT NO.
RECEIVING RESPONSE
REFERENCE SYSTEM
RECEIVING SYSTEM
DEPTH
WATER TEMPERATURE
DATE
TEST DISTANCE
FREQUENCY IN CYCLES PER SECOND
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM²

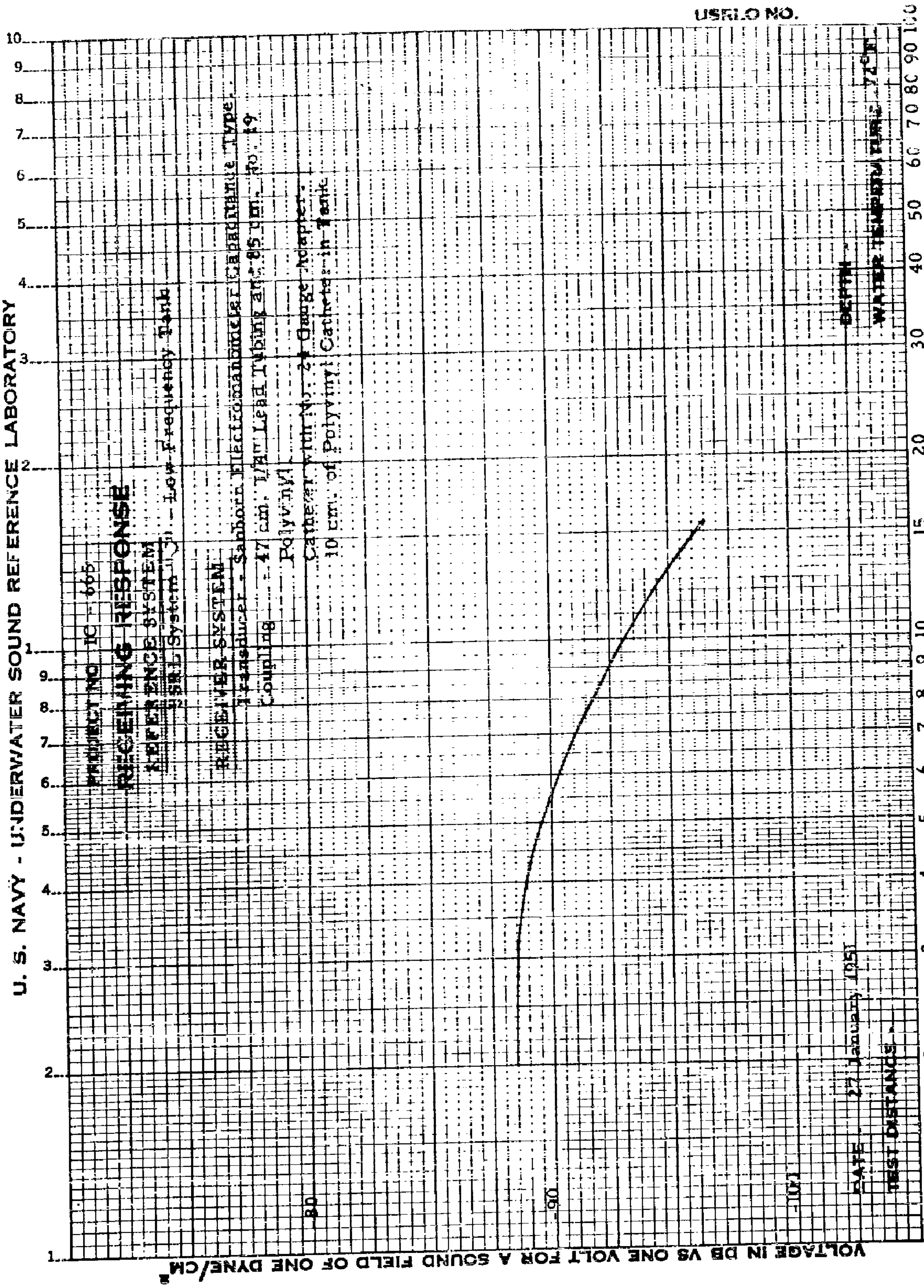

FREQUENCY CPS
3
10
30
100
300
1000
RELATIVE AMPLITUDE
30
20
10
0
Measured
Calculated

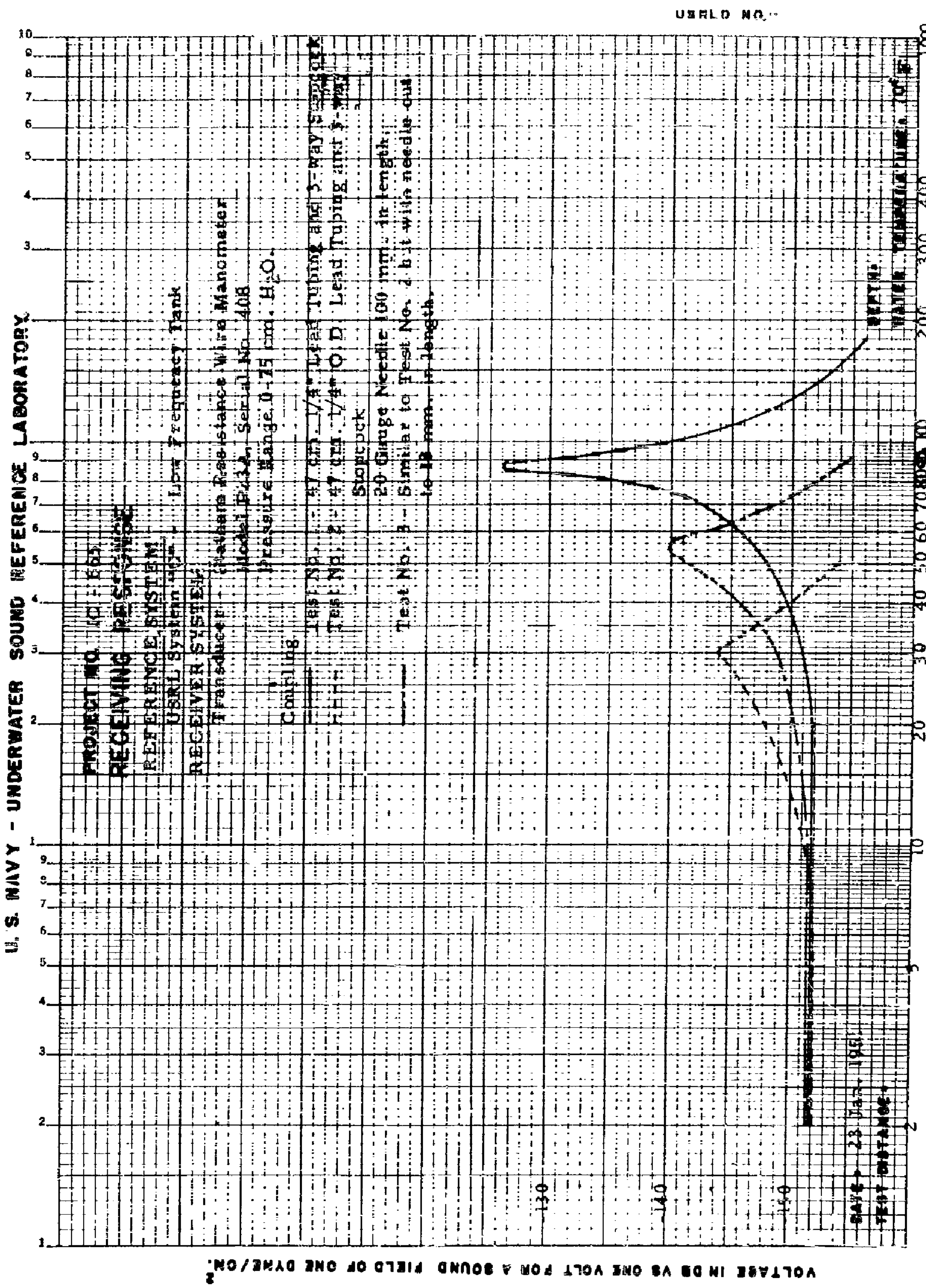

Pistonphone
USRL Low Frequency Tank
Step Function (Calculated)
FREQUENCY - CPS
RELATIVE AMPLITUDE db
-70
-80
-90
2
5
10
15
20
25
30
35
40
50
60
70
80
90
100

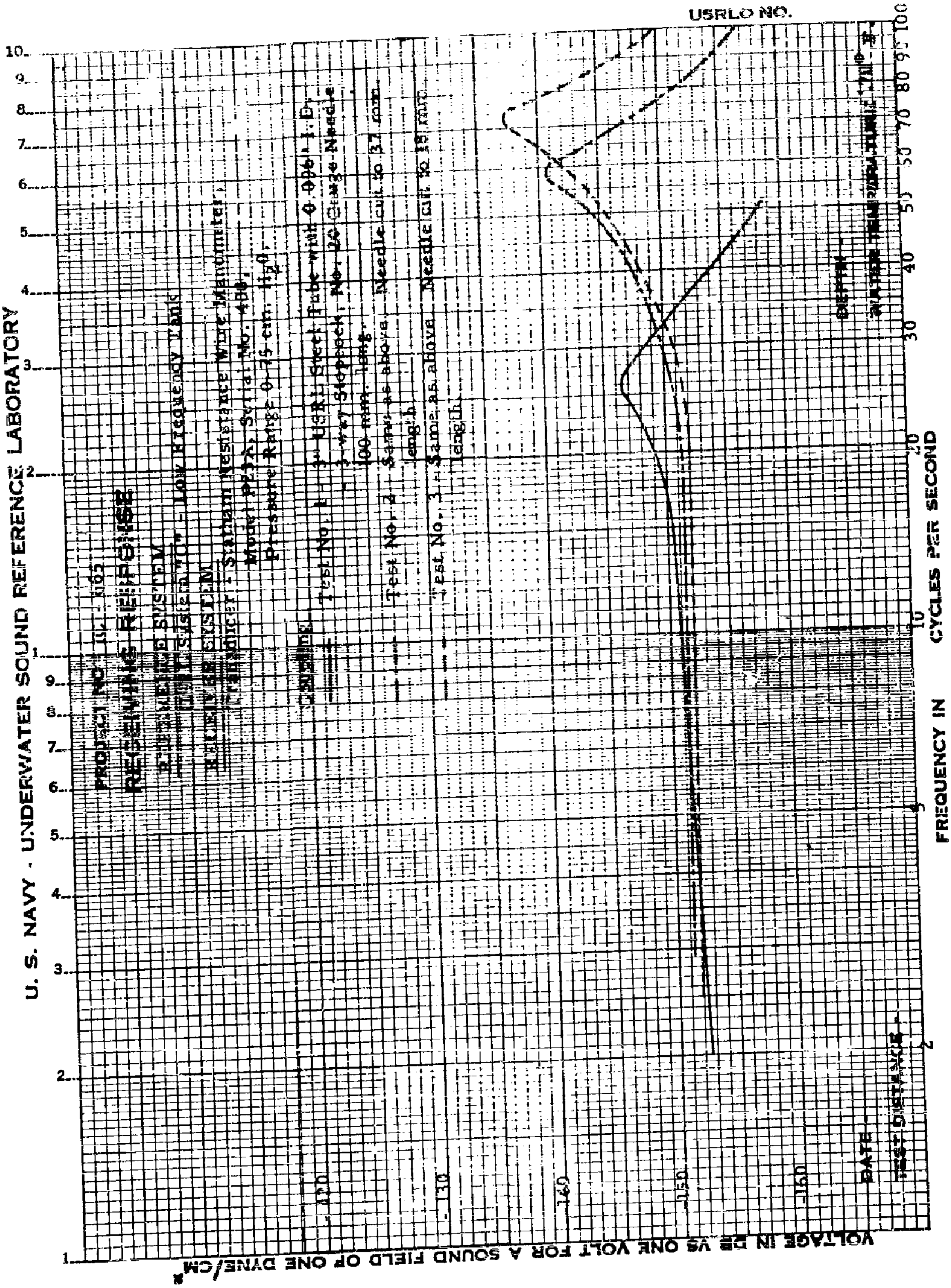

Project No. NM 001 060.07
RECEIVING RESPONSE
REFERENCE SYSTEM
AMAL Pistonphone
RECEIVER SYSTEM
Transducer - Lilly Capacitance Type
Pressure Manometer
Head No. 22
Coupling - 100 cm. No. 4 F Cournand
Cardiac Catheter With
3-way Stopcock
FREQUENCY IN CYCLES PER SECOND
10
100
1000
RELATIVE AMPLITUDE IN DECIBELS
3.0
2.5
2.0
1.5
1.0
.5
0

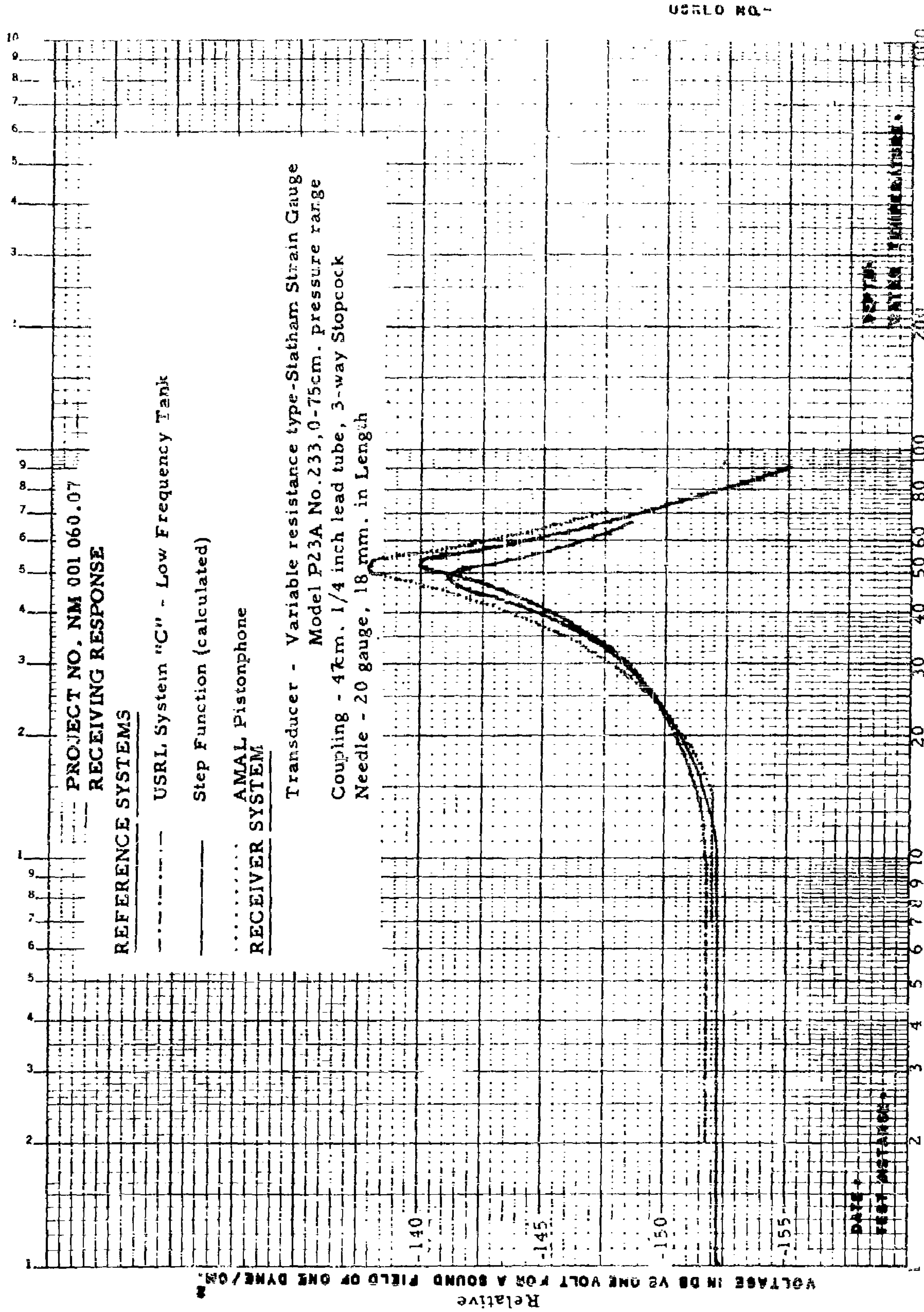

Project No. NM 001 060.07
RECEIVING RESPONSE
REFERENCE-SYSTEM
AMAL Pistonphone
RECEIVER SYSTEM
Transducer - Lilly Capacitance Type Pressure Manometer, Head No. 22
Coupling - 100 cm. No. 7 F Cournand Cardiac Catheter with 3-Way Stopcock
6.0
5.0
4.0
3.0
2.0
1.0
1
10
100
1000
Frequency in cycles per second
Relative Amplitude in Decibels

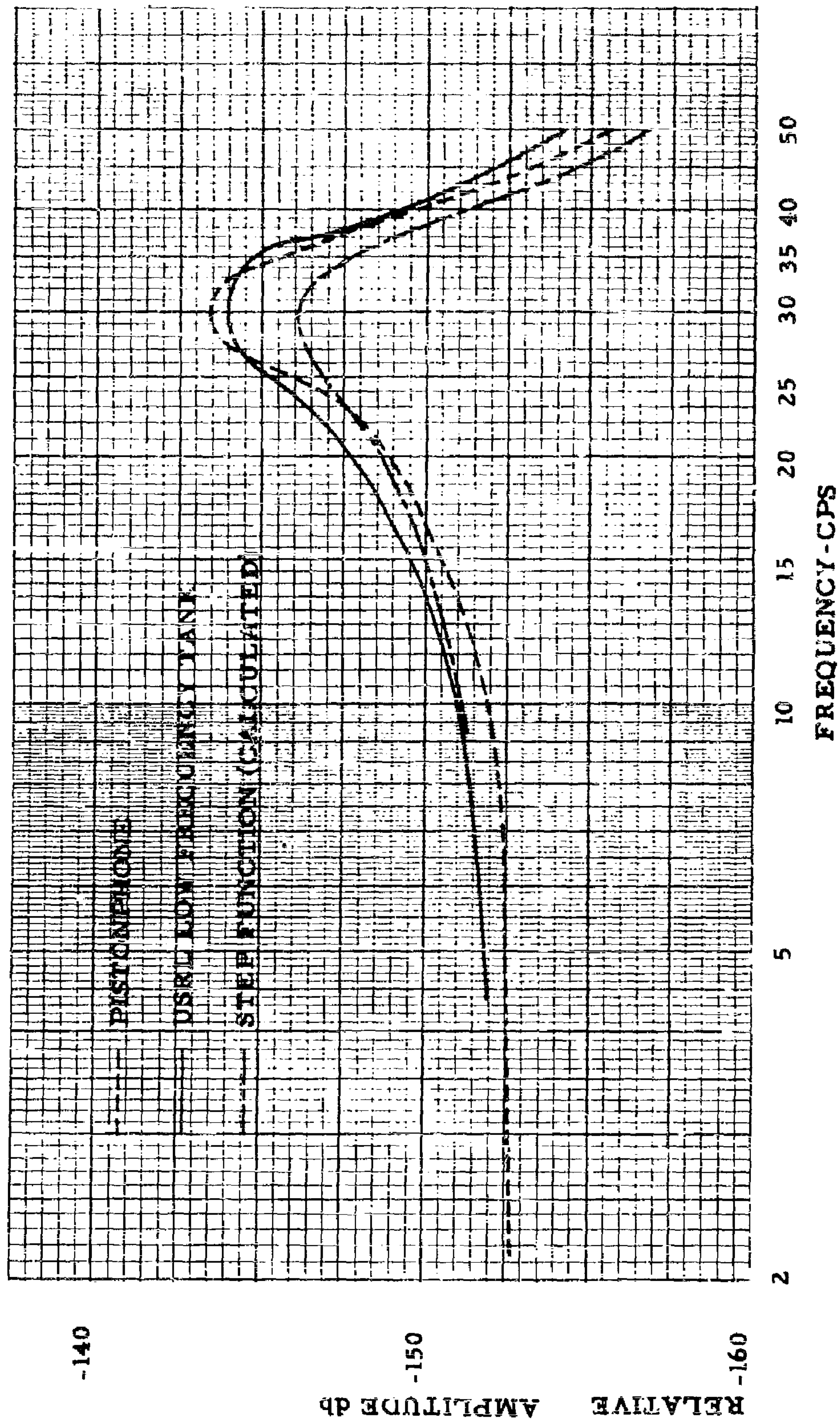

RELATIVE AMPLITUDE db
-80
-90
-100
FREQUENCY CPS
5
6
7
8
9
10
15
20
25
30
35
40
45
50
60
70
80
90
Pistonphone
USRL Low Frequency Tank
Step Function (Calculated)

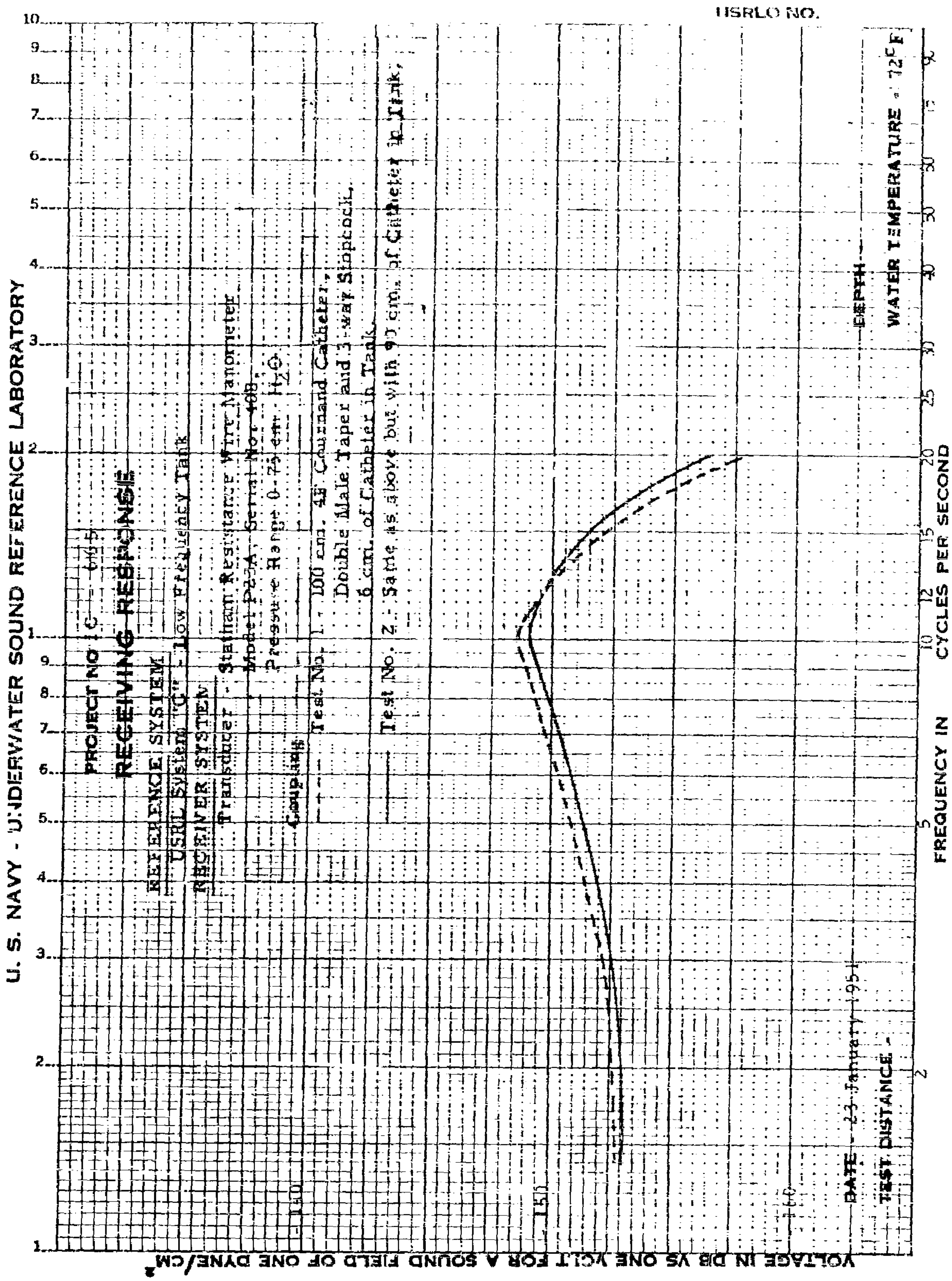

USRLO NO.
U. S. NAVY - UNDERWATER SOUND REFERENCE LABORATORY
PROJECT NO. IC - 665
RECEIVING RESPONSE
REFERENCE SYSTEM
USRL System 30 - Low-Frequency Tank
RECEIVER SYSTEM
Transducer - Sanborn Electromanometer Capacitance Type.
Coupling - 47 cm. 1/4" Lead Tubing and 85 cm. No. 19 Polyvinyl Catheter with No. 24 Gauge Adapter. 10 cm. of Polyvinyl Catheter in Tank
DEPTH
WATER TEMPERATURE 72°F
DATE 27 January 1951
TEST DISTANCE
FREQUENCY IN CYCLES PER SECOND
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM²
-80
-90
-100

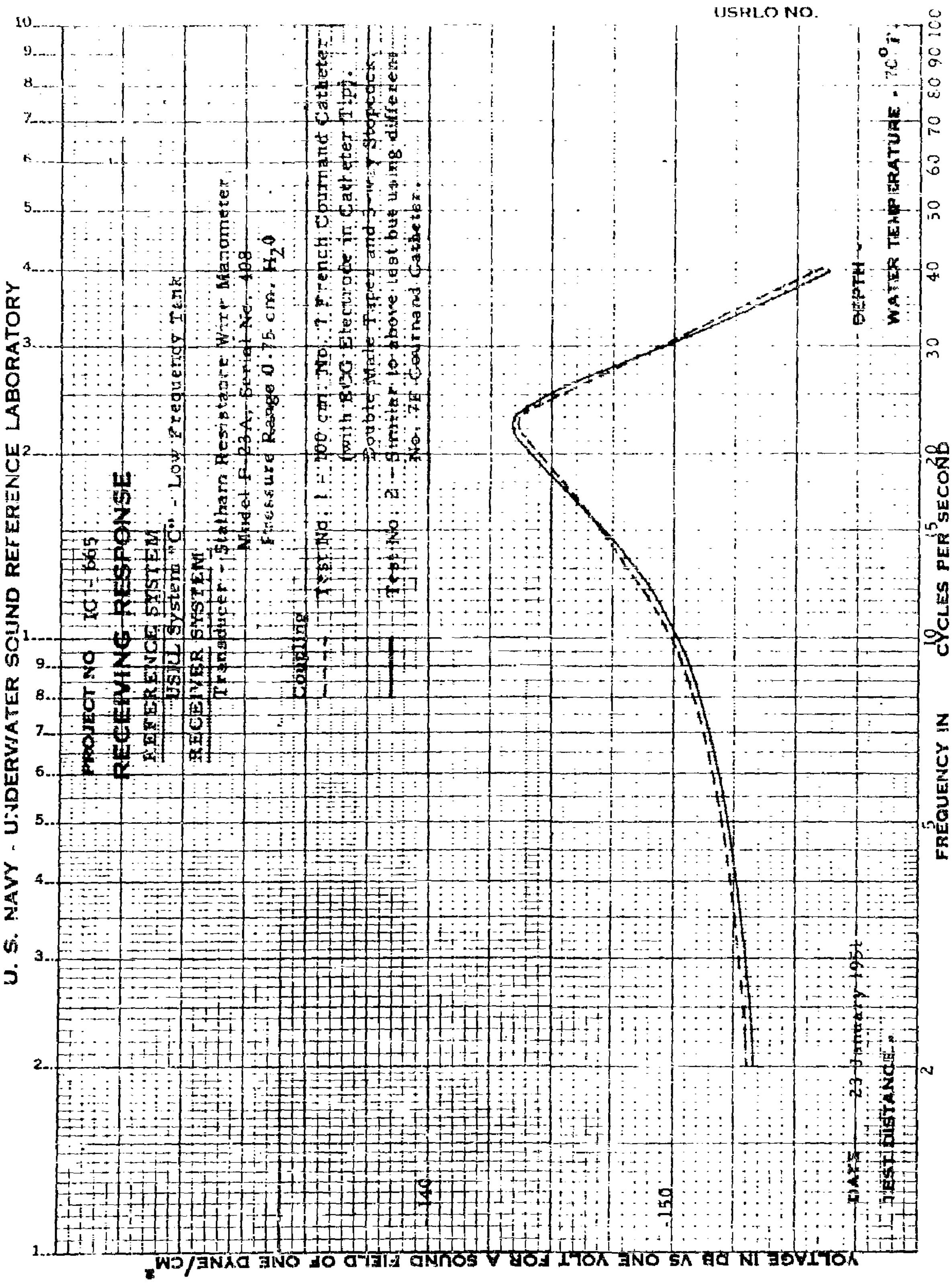

U. S. NAVY - UNDERWATER SOUND REFERENCE LABORATORY
PROJECT NO. IC - 565
RECEIVING RESPONSE
REFERENCE SYSTEM
USRL System "C" - Low Frequency Tank
RECEIVER SYSTEM
Transducer - Statham Resistance Wire Manometer
Model P23A, Serial No. 408
Pressure Range 0-75 cm. H_2O.
Coupling
Test No. 1 - 47 cm. 1/4" Lead Tubing and 3-way Stopcock
Test No. 2 - 47 cm. 1/4" O.D. Lead Tubing and 3-way Stopcock
20 Gauge Needle 100 mm. in length.
Test No. 3 - Similar to Test No. 2 but with needle cut to 18 mm. in length.
DATE - 23 Jan. 1951
TEST DISTANCE -
DEPTH -
WATER TEMPERATURE - 70°F
USRLD NO.
FREQUENCY IN CYCLES PER SECOND
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM.^{2}
130
140
150


U. S. NAVY - UNDERWATER SOUND REFERENCE LABORATORY
USRLO NO.
FREQUENCY IN CYCLES PER SECOND
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM²
RESPONSE
DEPTH
DATE
TEST DISTANCE


USRLO NO.-
PROJECT NO. NM 001 060.07
RECEIVING RESPONSE
REFERENCE SYSTEMS
USRL System "C" - Low Frequency Tank
Step Function (calculated)
AMAL Pistonphone
RECEIVER SYSTEM
Transducer - Variable resistance type-Statham Strain Gauge
Model P23A No.233, 0-75cm. pressure range
Coupling - 47cm. 1/4 inch lead tube, 3-way Stopcock
Needle - 20 gauge, 18 mm. in Length
FREQUENCY IN CYCLES PER SECOND
Relative
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM.2
-140
-145
-150
-155
2
3
4
5
6
7
8
9
10
20
30
40
50
60
80
100
200
1000
DATE
TEST DISTANCE
DEPTH
WATER TEMPERATURE


PISTONPHONE
USRL LOW FREQUENCY TANK
STEP FUNCTION (CALCULATED)
FREQUENCY-CPS
2
5
10
15
20
25
30
35
40
50
-140
-150
-160
RELATIVE AMPLITUDE db


U. S. NAVY - UNDERWATER SOUND REFERENCE LABORATORY
USRLO NO.
RECEIVING RESPONSE
REFERENCE SYSTEM
USRL System "C" - Low Frequency Tank
RECEIVER SYSTEM
Transducer - Statham Resistance Wire Manometer
Model P23A, Serial No. 408,
Pressure Range 0-75 cm. H2O
Coupling
Test No. 1 - 100 cm. 4F Cournand Catheter,
Double Male Taper and 3-way Stopcock,
6 cm. of Catheter in Tank
Test No. 2 - Same as above but with 90 cm. of Catheter in Tank.
DEPTH
WATER TEMPERATURE - 72°F
DATE - 23 January 1951
TEST DISTANCE -
FREQUENCY IN CYCLES PER SECOND
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM2


U. S. NAVY - UNDERWATER SOUND REFERENCE LABORATORY
USRLO NO.
PROJECT NO IC-665
RECEIVING RESPONSE
REFERENCE SYSTEM
USRL System "C" - Low Frequency Tank
RECEIVER SYSTEM
Transducer - Statham Resistance Wire Manometer
Model P-23A, Serial No. 408
Pressure Range 0.75 cm. H2O
Coupling
Test No. 1 - 100 cm. No. 7 French Cournand Catheter (with BDG Electrode in Catheter Tip).
Double Male Taper and 3-way Stopcock.
Test No. 2 - Similar to above test but using different No. 7F Cournand Catheter.
DEPTH -
WATER TEMPERATURE - °C °F
FREQUENCY IN CYCLES PER SECOND
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM²
DATE 23 January 1954
TEST DISTANCE
-140
-150

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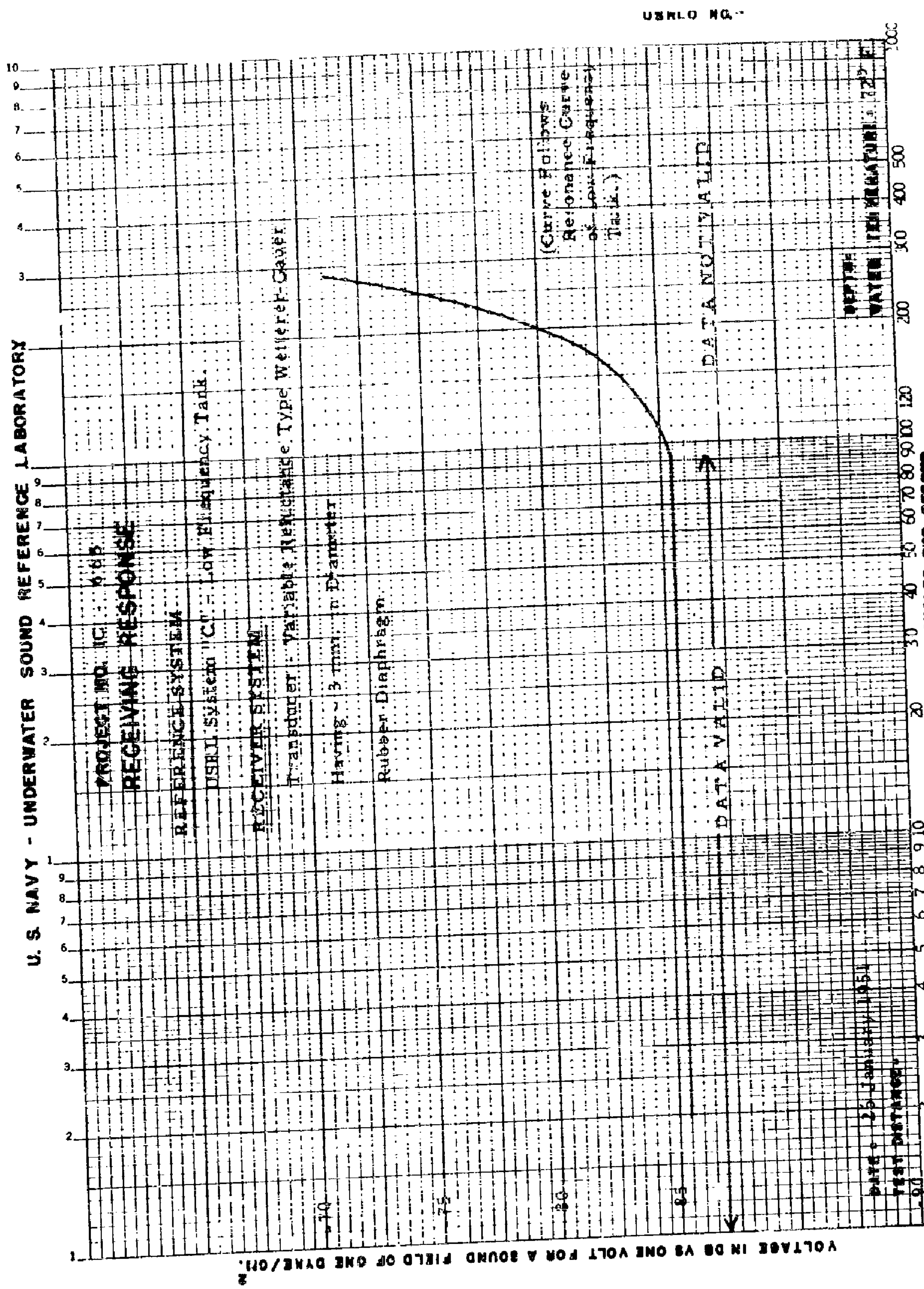
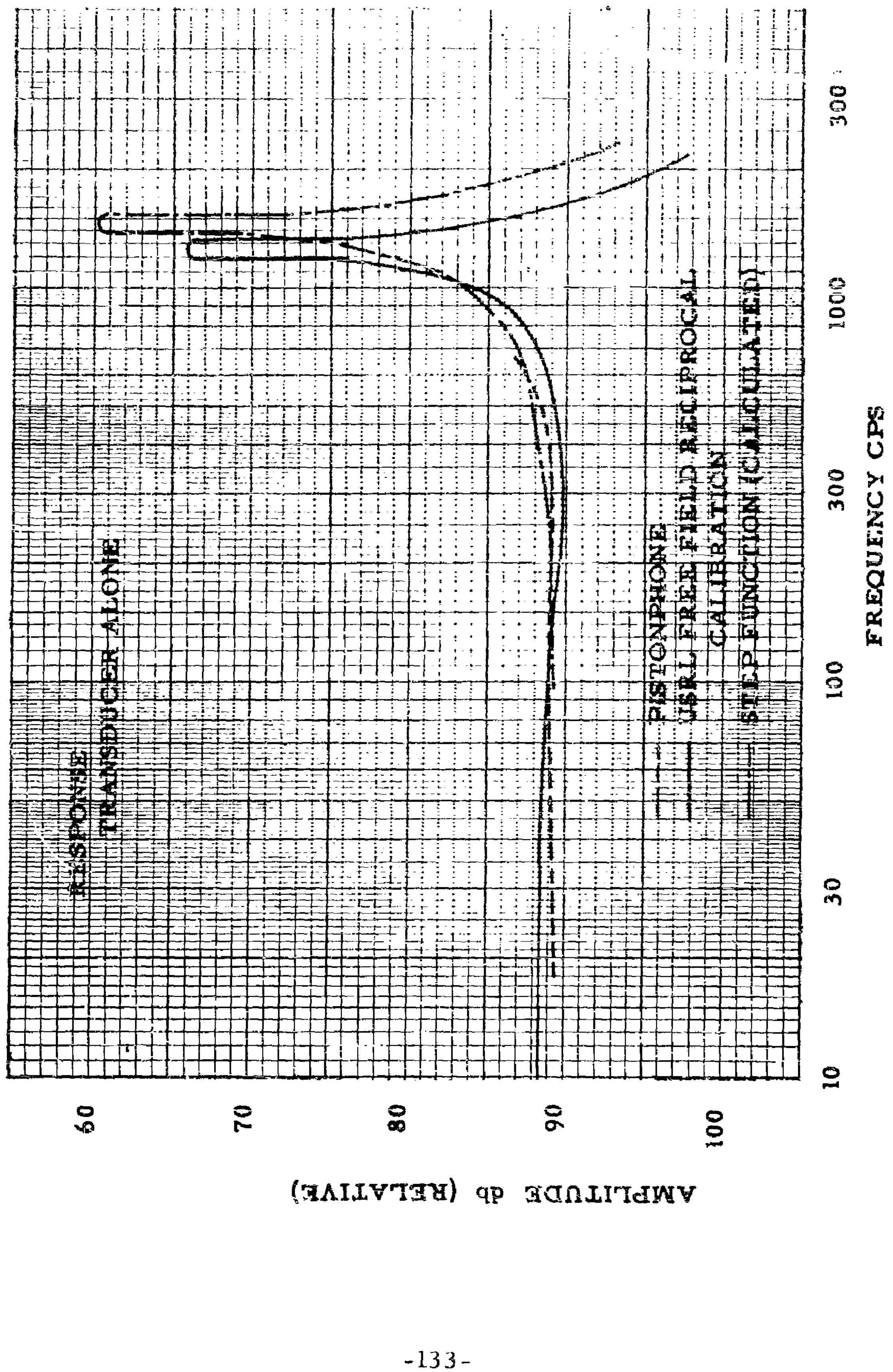
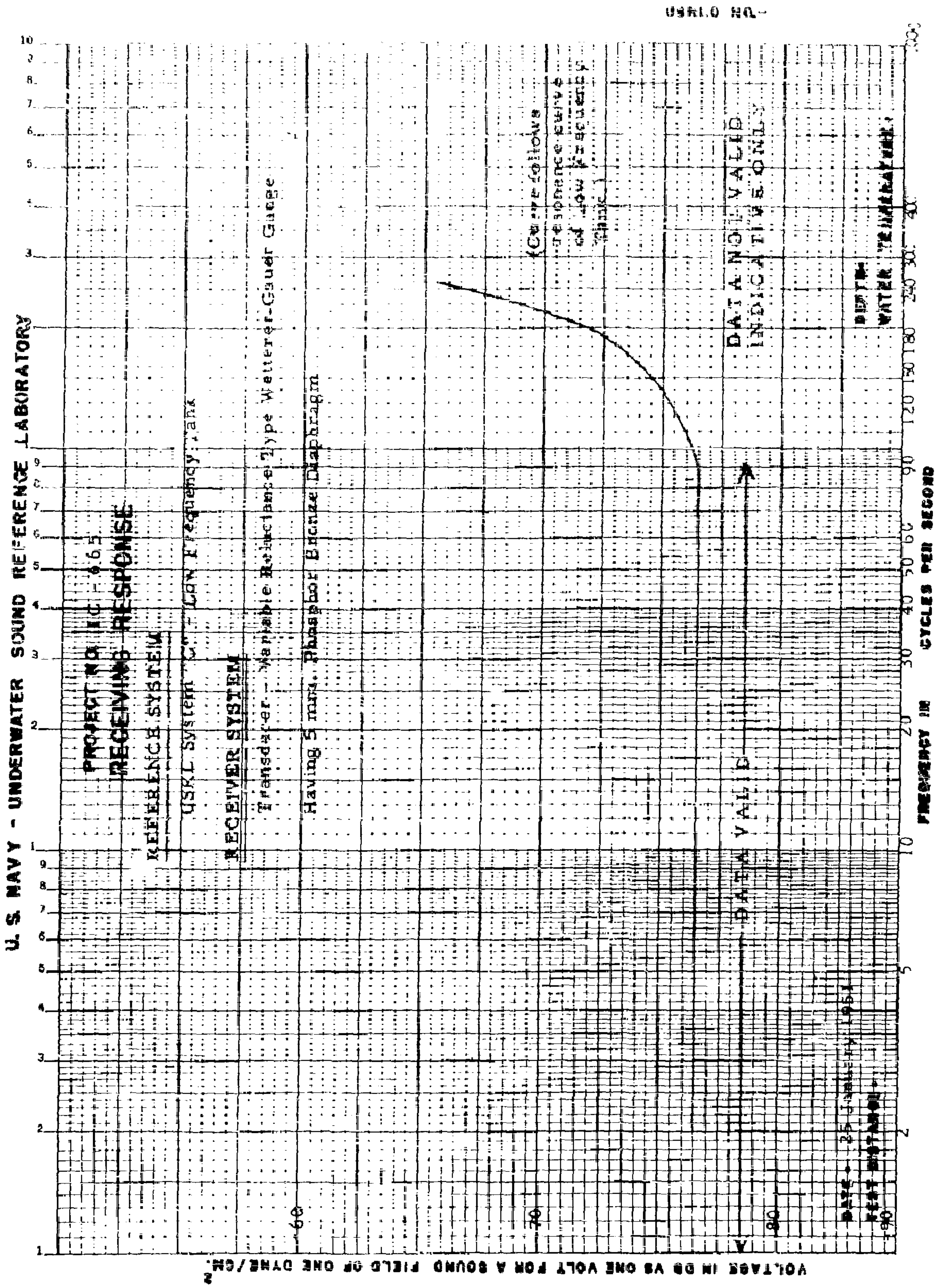
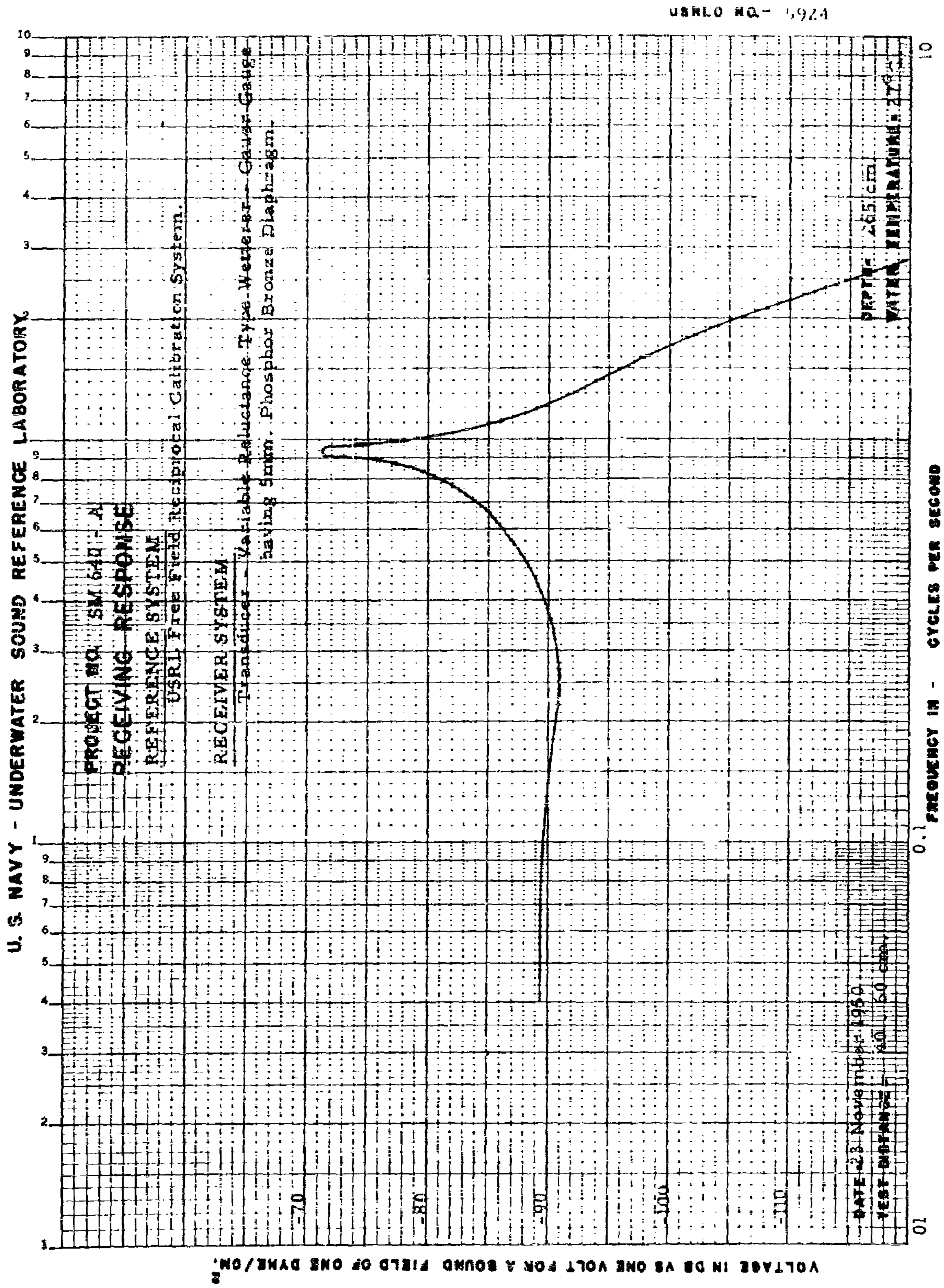



RESPONSE
TRANSDUCER ALONE
PISTONPHONE
USRL FREE FIELD RECIPROCAL
CALIBRATION
STEP FUNCTION (CALCULATED)
FREQUENCY CPS
10
30
100
300
1000
300
AMPLITUDE db (RELATIVE)
60
70
80
90
100


U. S. NAVY - UNDERWATER SOUND REFERENCE LABORATORY
RECEIVING RESPONSE
REFERENCE SYSTEM
RECEIVER SYSTEM
Having 5 mm. Phosphor Bronze Diaphragm
DATA NOT VALID
INDICATIVE ONLY
DATA VALID
FREQUENCY IN CYCLES PER SECOND
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM²


USRLO NO.- 5924
U. S. NAVY - UNDERWATER SOUND REFERENCE LABORATORY
PROJECT NO. SM 640 - A
RECEIVING RESPONSE
REFERENCE SYSTEM
USRL Free Field Reciprocal Calibration System.
RECEIVER SYSTEM
Transducer - Variable Reluctance Type Wetterer - Gauss Gauge
having 5mm. Phosphor Bronze Diaphragm.
DEPTH - 265 cm
DATE 23 November 1950
TEST DISTANCE 40, 50 cm.
-70
-80
-90
-100
-110
FREQUENCY IN - CYCLES PER SECOND
VOLTAGE IN DB VS ONE VOLT FOR A SOUND FIELD OF ONE DYNE/CM.²

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